THE FEMORAL NECK

An experimental study

of function, fracture mechanism,

and internal fixation

BY

VICTOR H. FRANKEL

UPPSALA 1960

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av

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BOKTRYCKERI AKTIEBOLAG

UPPSALA 1960

To my mother

and my wife

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AFHANDLING
Om
BEN-SJUKDOMAR.

Efter
Erkände Auctorers Arbeten
och Anmärkningar
Samlad
Af

ROLAND MARTIN, M. D.
Aſſeſſor i Kongl. Collegio Medico, Profeſſor Anatomiæ & Chirurgiæ, Ledamot af Kongl. Svenſka Wetenſkaps Academien och Chirurgiſka Societeten i Stockholm, ſamt Kongl. Götheborgſka Wetenſkaps och Witterhets Sällſkapet,

STOCKHOLM,
Tryckt hos Commiſſarien P. A. BRODIN, 1782.

Fractura Oſſis Femoris.

§. 1.

Detta ben är et ibland de ſvåreſta at inrätta och qvarhålla. Det fordras altſå förut en god kännedom om deſs ſtructur.

Fractura Colli femoris är nog ſå ſvår at hålla i ſkick, ſom obliqua, för de manga muſklar ſkull, ſom omgifva denſamma, och för obeqvämheten at komma åt at tilbinda denſamma.

INTRODUCTION

For the past quarter of a century the treatment of fractures of the proximal end of the femur by internal fixation has been pursued vigorously. At the present time, one might state that it has achieved world-wide recognition as the method of choice in the treatment of most fractures in this area. This recognition, however, has rightly not stilled the voice of criticism.

Research has been directed towards every facet of the fracture problem: statistical incidence, fracture mechanism, vascular supply, mechanical considerations, and many other subjects have been studied.

In this work, an attempt is made to study mechanical factors acting in the frontal plane, important to osteosyntheses, and related to osseous structure and metallurgical knowledge.

THESIS ON BONE DISEASES

This bone is sometimes one of the most difficult to reduce and to keep in position. It is necessary, therefore, to previously have a good knowledge of its structure.

A fracture of the femoral neck is at least as difficult to maintain in reduction as an oblique fracture on account of the many muscles which surround it and due to the fact that it is difficult to bind it.

Roland Martin, M.D. Stockholm 1782

CHAPTER 1

SURVEY OF THE LITERATURE

THE HISTORY OF INTERNAL FIXATION OF FEMORAL NECK FRACTURES

It is convenient to divide the history of internal fixation into two periods: the early one commencing in 1858 with von Langenbeck's attempt and the present period beginning in 1931 with the report of Smith-Petersen, Cave, and Vangorder.

In addition to von Langenbeck's (1878) case, treated by exposure of the fracture site and fixation with a nail passing into the femoral head through the greater trochanter, Koenig (1878) reported a similar operation. Lister (1880) performed the first fixation under aseptic conditions. The first series of cases was reported by Nicolaysen (1897) who treated the fractures of 13 patients with a steel spike which protruded through the skin and was used in conjunction with a hip spica.

Other types of fixation were sought in the early period. Delbet (1919) and Thomas (1921) reported their experiences with screws. Albee (1915) was one of many surgeons who used bone grafts. Hey Groves (1916) tried a four flanged nail.

Widespread interest followed the publication of the results of the cases treated by Smith-Petersen and his group in 1931. They used a three flanged nail which was designed to prevent rotation as well as vertical displacement of the femoral head. Although at first the nail was introduced by means of an open exposure of the fracture site, Johansson (1932 and 1934), Wescott (1932), and Henderson (1934) reported methods of introducing the nail 'blindly' using a cannulated nail and a guide wire.

With interest aroused in internal fixation of femoral neck fractures, numerous reports began to appear from many clinics describing other methods of fixation. Moore (1934), Telson and Ransohoff (1935), Sofield (1937), Compere (1940), and Harmon (1945) advocated multiple threaded wire fixation. Deyerle (1959) described a plate to hold multiple threaded pins.

The use of multiple spikes or nails was described by Gaenslen (1935), and Nyström (1944), Screws placed in a parallel manner were used by Geckeler (1937), von Bahr (1946), and Kinsella (1948), and a connected double screw was advocated by Schumpelick (1955).

Various forms of lag screws and cork screw bolts were developed by Jones

(1933), Henry (1934), Lippman (1936), Henderson (1938), Putti (1940), Lorenzo (1941) and Johnston (1944). Screws designed to exert continuous traction were produced by Greer (1938) and Virgin and MacAusland (1945). Stud and lock bolts were introduced by Cleary and Morrison (1940), Godoy-Moreira (1940), and Hardinge and Janzen (1950).

To prevent backing out of the appliances, nails were modified by means of cross pins by Pidcock (1936) and by cross screws by Thatcher (1943). Hart (1937) suggested the use of a wire bone suture. Side bars were applied by Jewett (1941), Moore (1944), Taylor, Neufeld, and Janzen (1944), Capener (1944), McKee (1944), McLaughlin (1947) among many others. Nails that could slide out of their side bars were designed by Pugh (1955), Luck (1956), and Massie (1956). A sliding screw-plate was described by Schumpelick and Jantzen (1953) and Wassner (1955), and a compression screw-plate was presented by Charnley (1956) and Charnley, Blockey and Purser (1957).

A flange was described by Cubbins, Callahan, and Scuderi (1939). A nail with a 'Z' cross section was developed by Thompson (1948) and a nail with a 'H' shape by Haboush (1954). A curved medullary nail was introduced by Lezius (1950) and a Y shaped appliance consisting of a medullary nail with a cross nail of an 'I' beam type extending into the head was described by Küntscher (1954). A 'Pistol' nail-plate consisting of a three flanged nail and a large plate held to the shaft with circumferential wires was discussed by Winkelbauer and Moser (1951). A nail-plate with a 'U' shaped nail and slotted holes in the plate was described by Rehbein (1948).

The possibility of fixation of the fracture site with appliances extending through the head into the acetabular wall was explored by Richard (1931) who used a screw; Bauer and Georg (1955) and Rosenthal and Sutro (1958) used two nails.

The trend over the years has been to develop stronger appliances, firm side bar fixation, and mechanisms to allow impaction and compression of the fracture fragments.

EARLIER INVESTIGATIONS ON THE MECHANICAL EFFICIENCY OF OSTEOSYNTHESES

Although internal fixation of the femoral neck has been practiced for over one hundred years, relatively few studies of the mechanical efficiency of the various appliances or of the strength of the osteosyntheses have been performed.

A paper describing the healing of experimental fractures of the hip in animals both with and without internal fixation was published by Senn (1881). Although

he did not deal with the mechanical principles involved, he established the point that the fractures healed, either by bone or strong fibrous union when internal fixation was used. Without internal fixation, there was much less tendency to heal.

Henry (1931) performed osteosynthesis on dogs and followed the course of healing and aseptic necrosis.

Kotrnetz (1933) tested the holding power of Smith-Petersen nails in femoral specimens which were subjected to vertical force acting on the head. The fractures were made by sawing through the neck. The head was found to slip from the neck almost immediately after force was applied and the osteosynthesis to disrupt completely between 50 and 100 kg.

Pauwels (1935) divided femoral fractures into three groups dependent on the inclination of the fracture line to the vertical. He also studied the mechanical factors important to an osteosynthesis.

Lehmann (1936) criticized the work of Kotrnetz as being performed on bones with smooth fracture surfaces, a condition not usually found in life. He performed similar loading experiments on specimens with differing fracture lines. One had a smooth fracture surface, this failed at 50 Kg. The others had fracture surfaces that were irregular and ran in different directions. The fractures that had more horizontal lines held together to above 130 Kg. The fracture with a vertical direction withstood 50–90 Kg before slipping took place. The nail bent at 90 Kg. He believed that the direction of the fracture line was very important to the strength of the osteosynthesis, as the more horizontally directed fracture lines allowed impaction of the bone and subsequent bony support.

Tinker and Tinker (1939) reported tests carried out on sawed and broken bones using Moore pins and Smith–Petersen nails. For the saw cut fractures the Moore pins provided stability up to 45 Kg of applied force. For bones that were broken, nailed, and impacted, up to 140 Kg was tolerated by the best pinnings. The Smith–Petersen nails were as good as the least efficient Moore pins.

Bader and Gui (1940) tested the Putti screw rather intensively. Tests on the screw alone showed that, depending where force was applied along the screw, it could stand perpendicularly applied forces of up to 145 Kg before breaking. They tested the ability of the screw to resist pulling out from the head of the femur and found the minimum pulling force to be 125 Kg and the maximum 475 Kg.

Femoral specimens were tested by first fracturing them with vertically directed force, which produced fractures with vertically directed fracture lines, and then reuniting the specimens with the screw. When these specimens were tested with vertical forces acting on the head of the femur several phenomena were observed. The head was forced down from the neck in three specimens at 60–70 Kg. It could be shown that the motion started almost simultaneously with the force. Other

specimens started to disrupt at between 125–375 Kg. They described ovalization of the screw hole in the neck due to bending and motion of the screw.

Godoy–Moreira (1940) reported the holding strength of a fractured femoral specimen containing a stud bolt. Under a vertical load of 100 Kg slight motion was detectible between the head and the neck. With a load of 150 Kg the sliding motion increased and was easily visible. When the load reached 251 Kg the screw began to bend and it broke under a load of 317 Kg.

Husted and Holst (1941) reported that, according to calculations and experiments, a four flanged nail was about 90 per cent stronger than a three flanged nail.

Compere, Wallace, and Lee (1942) published a series of experiments comparing the effectiveness of threaded wires, Steinman pins, and Smith–Petersen nails. Fractures of the neck of the femur were produced by vertically loading the head. The fractures were in most cases vertical ones. These fractures were then fixed with the chosen appliance and tested with a similar vertically directed force acting on the head. They found that the threaded wires could withstand 328 Kg of force if the wires were horizontally oriented, and 205 Kg if more vertically directed, on an average. The Smith–Petersen nails averaged 158 Kg and the Steinman pins 169 Kg. If the breakdown force for the osteosynthesis is compared to the force necessary to fracture the intact bone, the osteosyntheses produced between 21 and 27 per cent of normal vertical force resisting strength.

In testing the resistance to distraction, they found that it was necessary to have an average pull greater than 23 Kg for the Smith–Petersen nails, 40 Kg for the Steinman pins, and 53 Kg for the threaded wires.

Bohlman (1942) in a discussion of the results of Compere *et al.* stated that the breakdown force for fractures fixed with Moore pins was similar to that found for the threaded wires.

Compere and Wallace (1942) reported studies on aseptic necrosis in dogs. Their conclusion was that less necrosis occurred in the transcervical fractures treated immediately with accurate reduction and immobilization. A less satisfactory prognosis for head survival was found, if the reduction was not anatomically correct or the pinning of the fragments not adequate for complete immobilization.

Spotoft (1944) reviewed the problem of osteosynthesis from a mechanical standpoint. He performed experiments to test the strength of osteosynthetic materials and attempted to produce fractures of the femur.

The strength of nailed fractured specimens was tested by placing vertically directed force on the head of the femur, and by using a point on the head and another on the neck, to measure the distance the head slipped from the neck. Force was steadily increased by 20 Kg increments until a total of 250 Kg was

reached. The femur was then loaded up to 50,000 times at a rate of 30 loadings per minute, which was the equivalent of a 20 kilometer walk. Motion at the fracture site was noticed with forces as low as 50 Kg.

Spotoft also compared the results with Pauwels' grade I and Pauwels' grade III fractures and determined the differences between using a horizontally placed and a steeply placed nail. The findings were that a Pauwels' grade I fracture could take up to 490 Kg and Pauwels' grade III types 330 Kg. Using a Pauwels' grade III specimen with a steep nail and comparing it to a similar fracture with a horizontal nail, the steep nail seemed to be somewhat stronger, although this may have been due to the use of a side screw. The steep nail was found to withstand a force of 730 Kg and the horizontal nail 340 Kg before disruption took place, although at 250 Kg there was about 2.5 mm of slipping of the head on the neck.

Harmon, Baker, and Reno (1948) performed experiments on embalmed bones. Sawcut fractures were made across the midportion of the femoral neck at a 30 degree angle with the long axis of the shaft. Tests were made of the ability to withstand distraction and crushing forces. The endpoint of the crushing test was the earliest grossly detectable sudden yield or separation at the fracture site or elsewhere in the specimen. From their illustration of the loading tests it would seem that the fracture line was put under compression as the shaft of the femur was mounted at an angle of 45 degrees to the vertical, pointing the neck and head almost vertically. They tested twelve different appliances and found that multiple hexagon headed slender stainless steel screws yielded the highest values on resistance to both distraction and crushing.

The average breakdown force for osteosyntheses with Smith–Petersen nails was 362 Kg. The nail cut the cancellous bone and sometimes bent. Moore pin fixations withstood 245 Kg. The pins bent under load. Specimens with Moreira lag screws tolerated 270 Kg, bending of the screw occurred. The Lorenzo screw osteosyntheses withstood about 218 Kg; the screw cut the head. Specimens with a Lippman corkscrew bolt took 280 Kg. Osteosyntheses made with Compere threaded wires withstood 164 Kg; five stainless steel screws tolerated 360 Kg; and four stainless steel screws withstood 270 Kg. The shaft of the screws cut the cancellous bone of the neck and head.

Maatz and Lentz (1950) tested osteosyntheses made with a 'U' shaped nail in a horizontal and steep position. When the horizontal position was used, the head displaced on the neck under a 25–40 Kg load. The nail cut through the greater trochanter and neck spongiosa. The steep nail fixations took about 250–330 Kg of load before slipping of the head on the neck took place. Steel plates were interposed between the head and neck to rule out the effects of bony support at the fracture line.

Haboush (1952) investigated the stresses in cervical fractures of a two dimensional model of the proximal end of the femur with a photoelastic technique. A comparison was made between nails in horizontal and steep positions and nail-plates with high and low angles. He concluded that the nail-plate with the steep angle gave the best fixation, although this fixation depended on an intact lower cervical cortex. He calculated the stresses nails and nail-plates would undergo when a load was applied vertically to the head of the femur. He expressed the opinion that a nail-plate with a steep angle was best.

Martz (1956) reported results of experiments on fixation devices. He gave no definite endpoint for the test, other than failure. Moore and Jewett appliances failed with loads of 49 Kg. Neufeld, Thornton, and Pugh appliances failed in the range between 80–94 Kg. A Blount blade plate failed when loaded to 172 Kg. A fractured femur with a Smith–Petersen nail disrupted under 159 Kg of load.

Martz discussed Pauwels' conceptions of the forces acting on the femoral head. With these data and engineering studies showing the importance of corrosion fatigue to the strength of the appliance, he concluded that there should be a great factor of safety in any appliances. He expressed the opinion that appliances should be able to withstand forces in excess of 300 Kg.

Foster (1958) discussed trochanteric fractures of the femur treated by means of a McLaughlin four flanged nail-plate. He calculated the bending moment for failure of the various appliances using Martz's data. He believed that the bending moment that might be developed in a nail-plate junction in a fractured femur would be about 6000 kpmm.

Harvey, Hirsch, and Wilson (1959) reported a series of experiments in which stress-strain curves made by loading intact femurs were compared to the stress-strain diagrams from the fractured and nailed specimens. Various types of nails and screws were used. The conclusion was reached that the position of the appliance in relation to the cortex of the neck was of more importance than the type of fixation itself.

END RESULTS WITH CURRENTLY USED APPLIANCES

Medial Fractures

The results obtained with medial fractures seem to be determined in part at the time of injury. Evidence from venographic studies of the femoral head following fracture of the neck demonstrated that some heads were deprived of blood supply at the time of injury (Hulth 1956). All of these cases showed aseptic necrosis.

Two sets of criteria have been applied to the study of the results of medial

fractures. Palmer (1951) and Jensenius (1956) combined the rate of aseptic necrosis and non-union and used this as a criterion, as non-union in their opinion was not observed without necrosis. Jensenius tabulated the combined rate for the series of end-results reported by Linton (1944), Spotoft (1944), Carlquist (1947), Odén (1947), Sörlie (1954), and Nyström (1954), among others. The combined aseptic necrosis-pseudoarthrosis rate varied from 20–54 per cent when the Smith–Petersen nail was used. The rate with Nyström nails varied from 27–45 per cent.

In displaced intracapsular fractures, Cleveland and Fielding (1954) found a 22 per cent non-union rate and a 24 per cent aseptic necrosis rate. They observed, as did Spotoft (1944) among others, that the angle made by the fracture surface with the long axis of the bone did not seem to be a factor in union.

A further evaluation of the status of the patient was made by grading the result, using such terms as *good, fair,* and *poor*. The grading varied somewhat with the observer. Clawson (1957) reported a series of subcapital fractures: 39 per cent were satisfactory; the rest were bed, wheel chair ridden, or walked with crutches. Neer and McLaughlin (1948) found that the patient was satisfied in 86 per cent of the cases and the surgeon in 25 per cent, when the fracture was treated by open operation. Cleveland and Bailey (1950) found a completely satisfactory result in two out of three cases. Bailey (1956) reported good or excellent results in 42 per cent. Bado (1948) observed nearly identical results with a three flanged nail and the Godoy-Moreira bolt. Good results averaged slightly over 50 per cent for both.

There exist relatively few follow up studies carried on over a long period of time for the newer methods of osteosynthesis; that is, side bar fixation, steep nails, sliding nails, and compression screws. Statistics in small series of cases can be misleading, there can even be great diversity in the results between large series of cases treated with the same appliance.

Massie (1956) reported a non-union rate of 21 per cent using a nail at an angle of 155 degrees. Gelbke and Dziekan (1952) found a 12 per cent non-union rate and a 44 per cent aseptic necrosis rate with the steep Küntscher nail.

Jewett (1956) using side bar fixation, reported a non-union rate of 16 per cent and an aseptic necrosis rate of 10.5 per cent. He graded 87 per cent of the cases as excellent or good. Thirty per cent of the nails protruded through the head into the acetabulum. Eaton (1956) found clinical union in 18 out of 28 fractures in which a nail-plate was used. Protrusion of the nail occurred in 35 per cent.

Pugh (1955) found a 14 per cent non-union rate with a sliding nail. Luck (1956) reported 14 cases treated with a sliding nail. Union was delayed in three and aseptic necrosis had occurred in one, although the follow up time was not long. Massie (1959) reported his experiences with a steep sliding nail. The union rate

was 90 per cent and the aseptic necrosis rate 14 per cent. Of interest in this series, in which a steep sliding nail was compared to an ordinary steep nail, was that the cases of aseptic necrosis, although lower in frequency, were more severe in the sliding nail series.

Charnley (1959) stated that an 85 per cent clinical success rate was possible one year after operation when the compression screw-plate was used in displaced fractures.

Attempts have been made to influence the rate of healing of fractures and the aseptic necrosis rate by changing the mechanics of the fracture line through an osteotomy, first popularized by Pauwels (1935). Grafts have also been used to enhance healing. Results here, however, have not been better than simple nailing. King (1950) observed union in 69 per cent with nailing, 71 per cent with nailing and grafting, and 72 per cent with osteotomy. DePalma (1950) found a 10 per cent non-union rate and an 18 per cent aseptic necrosis rate in a series of 40 transcervical fractures treated by wedge osteotomy. Hudson, Bartels, and Freese (1951) reported that 22 cases were good, five were fair, and 14 were poor in a series of 41 cases of Pauwels' grade III fractures treated by McMurray osteotomy. Patrick (1949) found non-union in 13 per cent of 47 cases treated with a graft and nail. Aseptic necrosis occurred in four cases and osteoarthritis due to graft penetration in 21 per cent.

In summation, with current methods of osteosynthesis of medial fractures, healing takes place in about 80–90 per cent, aseptic necrosis then occurs in roughly 25 per cent of the healed cases. In follow up examinations, about one-half of the results are classified as good or satisfactory. Results have not been noticeably improved by primary osteotomy or grafting.

Lateral Fractures

In contrast to medial fractures, the major difficulties found in the treatment of lateral fractures were due to secondary displacements, coxa vara, protrusion of nails, and disruptions and bending of the fixation apparatus. The residual deformity rate varied from 8 to over 90 per cent depending on the criteria used in series of cases reported by E. M. Evans (1951), Taylor, Neufeld, and Nickel (1955), Parker (1955), Robey (1956), Olshousen and Miles (1957), and Wade, Campbell, and Keren (1959).

E. M. Evans (1949) divided intertrochanteric fractures into stable and unstable types depending on the degree of injury to the cortex of the calcar (inferior cortex of the neck). The fracture was stable when the cortical bone was intact. When the inferior cortex was involved in the fracture, there was a tendency to coxa vara

whether or not the fracture was nailed or treated by traction methods. He reported (1951) that 25 per cent of the trochanteric fractures were of the unstable type and some degree of coxa vara was expected. Using the Capener nail-plate, he found that the more severe deformities were associated with bending of nails or fracture of the appliance. Cram (1955) also called attention to the unstable fractures.

STRUCTURE AND FUNCTION OF THE PROXIMAL FEMUR

Basic to the problem of osteosynthesis of a fracture of the femoral neck is a knowledge of the forces acting normally on this area. This information is helpful if calculations are to be made of the stresses and strains to which osteosynthetic appliances might be subjected. Because of the seemingly close correlation between structure and function of the proximal end of the femur, these two facets of the over-all problem have been considered together.

Internal Anatomy of the Femoral Head and Neck

Interest in the proximal end of the femur as a weight bearing structure arose following the description of the inner architecture of the femoral neck by Bourgery (1832) and Ward (1838). Ward's conception of the arrangement of the trabeculae was that of a street lamp suspended from a pole. Both treatises promulgated the theory that the direction of the trabeculae had something to do with the direction of weight bearing.

Wyman (1849) regarded the shell of the neck as a bent tube adapted to resist pressure by its oval form. The walls of this bony tube were further supported by the disposition of the cancellous bone, which acted as so many braces within. The trabecular systems were also thought to form part of an arch and to support directly a portion of the weight of the body and transmit it to the walls of the neck. Humphrey (1858) noted the similarity of the cancellous structure of the neck to a series of arches. He found a gradual disappearance of this structure with age which he thought accounted for the fracturability in this group. Wagstaffe (1874) stressed the importance of the inner (medial) wall of the neck and shaft.

The next advance was brought about through the collaboration of von Meyer, an anatomist (1867) and Culman, an engineer. They compared the femoral head and neck to a common engineering structure—the Fairborne crane. The conclusion reached was that the bony trabeculae were laid down along the lines of maximal internal stress. This analysis was the basis of the trajectorial theory of bone form. According to this theory, spongy bone trabeculae follow lines of maximal internal

stress, cross each other at right angles, and are perpendicular to the surface of the bone or articular cartilage. Some of these were thought to resist compression and others tension. This work was the basis used by Wolff (1870 and 1892) in his formulation of the laws of bone growth. This law implies that every change in the form and function of a bone produces changes in the trabecular architecture and external form in accordance with mechanical laws.

Many criticisms have been made of the trajectorial theory. Among the criticisms were that the line of force used in the original study was parallel to the shaft, the femoral shape was different from the model, a solid homogeneous material was used in the analysis (Koch 1917), and the trabeculae only resembled tension lines of a crane when the femur was sectioned frontally (Küntscher 1934, 1935 a, and 1936). Jansen (1920), found areas in which trabecular architecture did not agree with the trajectorial theory. He believed that the jerking pressure of muscles combined with the pressure of gravity was the chief mechanical stimuli for bone formation.

Carey (1929), in his analysis of muscle action and bone thought that the 'back pressure vectors' of the muscles acting on the joints was the main cause for bone architecture. In the frontal section of the femur, he thought that the trabecular system extending from the femoral head to the lower aspect of the neck was produced by back pressure vectors of the flexor, adductor, and medial rotator muscles of the hip joint. The trabeculae present in the superior portion of the neck were produced by the pressure vectors of the extensor, abductor, and lateral rotator muscles of the hip joint. Evans (1957) criticized the back pressure theory of bone formation as failing to completely describe the vector, as Carey only gave the direction and not the magnitude of the vector. Furthermore, the direction and magnitude of the back pressure vector was constantly being changed during movement at the joint because the angle of application of the muscle with respect to the moving bone was consequently changing.

Calculations have been used by Inman (1947) and Backman (1957) to show that the medial trabecular system is in line with the normal load direction on the femoral head.

In addition to speculations and calculations based on the structure of the trabecular systems as seen on frontal projection of the femoral head and neck, another structure seen best on sagittal sections through the femoral neck has received attention. This structure, first described and named *Schenkelsporn* by Merkel (1874), is a compact area of cancellous bone in the neck. It is known in the English literature as the *calcar femorale* and has recently been described anew by Harty (1957) who deliminated it as "... a dense vertical plate of bone within the femur originating in the posteromedial portion of the femoral shaft, under the

lesser trochanter and radiates laterally through the cancellous tissue toward the greater trochanter". In common usage the term *calcar femorale* has incorrectly come to mean the thickened inferior cortex of the femoral neck.

Carey (loc cit.) thought that the calcar femorale was due to the back pressure vectors of the hip flexors and extensors. Analysis today in light of the strong muscular forces involved in pelvic equilibrium would make it seem that this structure was placed so as to resist bending stresses on the femoral neck and could conceivably be the result of back pressure vectors of the abductors and adductors.

Farkas, Wilson, and Hayner (1948) dissected out a highly independent laminar bony system in the area of the calcar femorale and the medial trabecular system. They believed that it is a compression system, resorbed in its distal part with age, damaged during adduction fractures, and supplying blood vessels to the fracture site.

Dixon (1910) in his analysis of the femoral neck, observed a structure—the *lamina femoralis interna*. This was described as a tubular surface beneath the trochanters, made up of dense cancellous bone. The lamina were, he believed, arranged in a spiral manner around the inner surface of the cortical layers, adding greatly to be strength of the neck.

Calculations of the forces Acting on the Femoral Head

Calculations were performed by a number of authors, among them Fick (1904, 1911), Strasser (1908, 1917), and Grunewald (1912, 1920), showing that muscles acting over a joint produced force on the articulating bones. Studies by Grunewald (loc. cit.), Koch (1917), Marique (1945), and Backman (1957) were made on the stresses produced in the upper end of the femur by a load on the femoral head.

Braune and Fischer (1889) and Fischer (1899, 1900) studied the forces involved in gait. They adopted the idea of chronophotometry (Marey 1882) and developed it as a research tool. This technique consisted of attaching Geissler tubes to various portions of the body and photographing the moving subject as the lights flashed, allowing determination to be made of the velocities and accelerations of the joints in three dimensions. They studied the center of gravity of the extremities, and the inertias for each part of the body. Earlier, Borelli (1679) had determined the center of gravity of the body. Much of the foregoing work has been reviewed by Steindler (1955) and the pathological implications explored.

With this work as a basis, Pauwels (1935) calculated the forces acting on the head of the femur under several different conditions. He stated that on quiet two leg stance, ignoring muscular and ligamentous factors, the weight on each femoral head is one-half of the superimposed weight. When standing on one leg, the force

on the head is about three times body weight; and when walking, the support limb may have a load of five to six times body weight acting on the head of the femur. The effects of the antalgic gait and the use of a cane on the force on the head were noted. Inman (1947) and Backman (1957) have reported similar studies.

Inman (loc. cit.) calculated the torque about the hip joint. This value was checked experimentally by means of electromyographic studies. He compared the electromyographic values for abductor force to known values of hip torque produced by having his subjects pull against a spring. He then compared the electromyographic values produced in the muscles on standing on one leg to the calibrated values. The pull of the individual muscles and tendinous structures was calculated. Finally, the force acting on the head of the femur and its direction were calculated by means of vectorial analysis. He concluded that the minimum static pressure on the head of the femur on standing on one leg was approximately 2.4 to 2.6 times body weight. Determinations were made of the relative importance of the abductor muscles and the ilio-tibial tract in maintaining a level pelvis, and was found to be equally divided between the two structures. When the pelvis sagged 15 degrees, the fascial tension resisted almost all of the torque. When the pelvis was elevated 15 degrees the torque was resisted by the muscles alone.

Backman (loc. cit.) recalculated the forces acting on the femoral head and the state of stress occurring in a cross section through the neck of the femur. He gave conditions for equilibrium of the pelvis when standing on one leg, and by substituting data reported by Pauwels into his equations, found that the static pressure on the head is 2.92 times the body weight acting in a direction 15 degrees from the vertical. Backman has also described a plane of symmetry in the femoral head and neck named the 'principal plane'. He believed that the load direction coincided with this plane. He calculated the forces acting in the principal plane in a cross section of the neck. From these calculations a graph was drawn giving the stresses in various parts of the femoral neck under different load directions.

Blount (1956) called attention to the importance of these studies in the treatment of various disorders of the hip, with particular emphasis on the use of a cane. Denham (1959) demonstrated graphically the manner in which various types of operations around the hip joint alter the forces acting on the femoral head.

Some indication of the pressure produced in the hip joint by the combined effect of the body weight and abductor pull has been produced by the experiments of Osborne and Fahrni (1950). Using a pelvic preparation consisting of a pelvis with an attached femur, with the hip joint containing a fluid filled rubber bag connected to a manometer, they measured the pressure passing through the joint when various superimposed weights were balanced by gluteal pull. In the positions of abduction, mid-position, and adduction of the hip joint, the joint pressure in

pounds per square inch seemed to be the summation of the body weight and the gluteal pull. The direction of the resultant force was given by means of vectorial diagrams, but was not measured, nor was the distribution of the load on various parts of the head shown.

Studies of Forces in the Femoral Neck

Several standard engineering techniques have been applied to the study of the femoral neck by various authors. These experiments have shown that the femur is an elastic body and follows Hook's Law; that is, stress is proportional to strain, within certain limits. The studies also showed that muscular and ligamentous forces have an effect on the bone and the distribution of strain in the bone. Some of the more important techniques are summarized below.

Küntscher (1934, 1935 a, 1935 b) used a colophonium coating which, after drying on the bone, cracked on tensile strain. The cracks were transverse to the direction of the tensile strain and were concentrated at the sites of highest strain. Evans *et. al.* (1948 a, 1948 b, 1952, 1957) used the Stresscoat method in similar studies. Stresscoat is a brittle resinous lacquer which also cracks when compressive or tensile strains are applied. The sites of maximum strain concentration can be determined by this method. A method of calibrating the lacquer could be used.

Mechanical devices for measuring small deformations were applied to the study of stress and strain in bones by Küntscher (1936) and Marique (1945).

Strain gages are based on the principle that changing the dimensions of a wire alters its electrical resistance. When glued to a test object, the alteration in the electrical resistance due to change in the length of the wire may be measured by means of a bridge. Hirsch and Brodetti (1956 a and b) have used this method to follow compressive and tensile strains in the neck of the femur and have shown that it follows Hook's Law.

Plastic models of the proximal end of the femur were made by Milch (1940) and Pauwels (1948, 1949) and transilluminated with polarized light. When load was applied, light and dark bands appeared which showed the stresses in the model. They have used this technique to show the effect of muscle forces and ligamentous action on the femur. It was demonstrated that forces simulating the abductors and iliotibial tract change the stress pattern produced in the femoral proximal end by vertical loading. A criticism of this method is that the isotropic plastic model is unlike bone.

Jakobsson (1954) used a plane model of the pelvis and proximal femur to study the distribution of stress in normal and dysplastic 'hip joints'.

The Ultimate Strength of the Femoral Neck

The ultimate strength in kilograms is the maximum load acting on the femoral head that the neck is able to support. Table 1 contains data from the literature of the ultimate strength of the femoral neck and the type of fracture produced. The specimens were not classified by sex and age in most studies. The loads used were static, and the femoral shafts were held in a vertical position.

Table 1. *The ultimate strength of the femoral neck.*
(Static force parallel to the femoral shaft).

Author	Date	Type of Bone	Fracture Force	Type of Fracture
Messerer	1880	Fresh femurs	♂ 700–1075 Kg ♀ 425– 600	Mainly transcervical
Dixon	1910	Fresh femurs	818–1130	Mainly transcervical
Kolodny	1925	Fresh femurs	♂ 545– 590 ♀ 445– 500	Mostly transcervical
Gaenslen	1936	Embalmed femurs and pelves	590–1360	Majority transcervical
Compere, Wallace, Lee	1942	Fresh femurs	256–1252	Mostly transcervical
Spotoft	1944	Fresh femurs	♂ 1170–1430 ♀ 600–1010	Mostly intertrochanteric
Smith	1953	Embalmed femurs and pelves	440–1910	All types
Hirsch, Brodetti	1956	Fresh femurs	325– 870	Mostly transcervical

In addition to the authors mentioned in Table 1, the mechanism of fracture of the femoral neck was investigated by Riedinger (1874), Kocher (1896), Frangenheim (1906), Duhamel (1947), Spears and Owen (1949), Evans, Pederson and Lissner (1951), Backman (1957), and Hirsch and Frankel (1960). These authors produced fractures using forces causing compression, traction, bending, torsion, and combinations of the same.

The weight bearing strength of bone in the head and neck was studied by Göcke (1933) who found that the strongest part of the neck was the lower distal portion. Nyström (1938) stated that the center of the head was the strongest area for head sections. Hardinge (1949) found the area of greatest strength to extend along a tract between the medial end of the inferior cortex of the neck and the middle of the superior surface of the head. Maatz and Lentz (1950) noted that the center of the head resisted crushing by a 'U' shaped nail until a force of 180–290 Kg was exceeded.

CHAPTER 2

THE LOAD BORNE BY THE FEMORAL HEAD

The methods used in the past to reach an understanding of the loads on the head of the femur during various phases of locomotion have been to assume that the human body is capable of being analyzed according to the laws of mechanics.

DERIVATION OF THE FEMORAL HEAD LOAD BY LAWS OF MECHANICS

The assumption is usually made that the pelvis is balanced on the head of the femur as a lever system, with the head of the femur acting as the fulcrum, the body weight and opposing muscle forces being the load and the force. Although forces are acting in space around the head of the femur, for purposes of analysis in this paper, only those components acting in the frontal plane are considered.

If the center of gravity of the body is displaced to one side or other of the fulcrum; that is, the center of the head of the femur, the pelvis will tend to fall to that side unless opposed by a moment equal to the moment causing the rotation. The head of the femur is assumed to be globular, with the center of the head corresponding to the rotation of the hip joint. In the subject standing on one leg with the pelvis held horizontally, either at rest or during locomotion, a moment develops tending to depress the pelvis. This moment is dependent on the body weight and on the distance between the center of gravity and the center of the head. To hold the pelvis level, a similar moment must be developed on the opposite side of the fulcrum (fig. 1). As the two moments must be equal, the equilibrium will be:

Equation 1 $$G \cdot b = M_v \cdot c$$ (single leg support) (horizontal pelvis)

G is the body weight.

b is the distance between the gravitational force from the body and the center of the femoral head, which will be equal to one-half the distance between the femoral head centers.

M_v is a force vector. It must be emphasized that M_v refers to all muscle and ligamentous action over the hip joint occurring simultaneously, and not to the abductor group alone.

It is assumed that the weight of a lower extremity is one-sixth of the body weight, and the gravitational force line from the free leg passes through the center of its femoral head.

The two forces, G and M_v, will have a reaction, R_v, acting on the head of the femur. To find R_v it is necessary to substitute in the following equations, calculated from the equilibrium condition.

Equation 2 $$R_v \sin \beta = M_v \sin \alpha$$ (single leg support) (horizontal pelvis)

Equation 3 $$R_v \cos \beta - M_v \cos \alpha = \frac{5}{6} G$$

Solving for R_v:

Equation 4 $$R_v = \frac{b}{c} G \cdot \frac{\sin \alpha}{\sin \beta}$$

Solving for β:

Equation 5 $$tn\beta = \frac{\sin \alpha}{\frac{5}{6} \cdot \frac{c}{b} + \cos \alpha}$$

α is the angle M_v makes with the vertical
β is the angle R_v makes with the vertical

With the exception of G and b, these quantities do not lend themselves to direct measurement. To find c, angle α, and M_v it is necessary to determine which muscle forces are acting, in what strength, and with which action line. It is also necessary to have the same knowledge in regard to ligaments. So far, no known practical way of testing the force produced by a muscle in situ has been developed. Attempting to calibrate the electromyogram may lead to erroneous data due to the fact that the electrical changes are not directly proportional to the contraction process (Eberhardt *et al.* 1947). The electromyogram is useful, however, in determining qualitatively the muscle groups acting simultaneously. The action line of a muscle will vary depending on what part of the muscle happens to be contracting.

Chronophotometry is an indirect method of calculating dynamic forces by measuring the velocities and accelerations of body parts. This method depends for its evaluation on a knowledge of the masses and centers of gravity of the body and its parts, information that, for the most part, must be gained from cadaveric study.

As R_v at present cannot be measured, it must be calculated from equation 4 which is illustrated in figure 2 where R_v/G is shown as a function of the ratio c/b

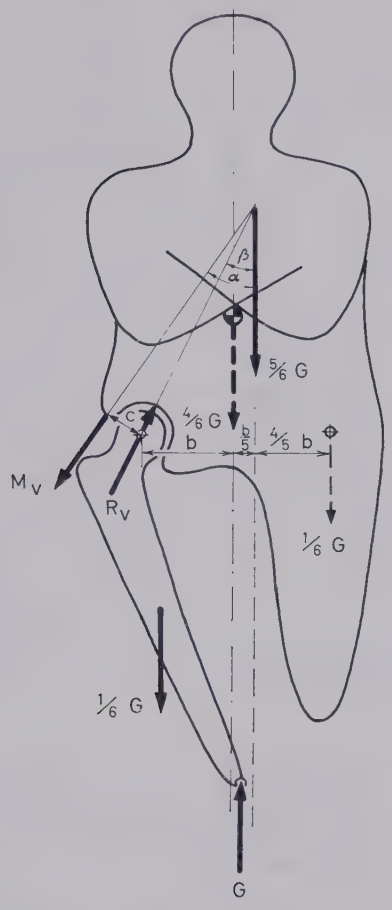


Fig. 1. The forces acting on the femoral head in the frontal plane. One leg stance and horizontal pelvis. G, body weight; M_v, resultant of musculo-ligamentous forces; α, angle between application of musculo-ligamentous action and the vertical; R_v, resultant of forces acting on the femoral head; β, the angle between the resultant of forces and the vertical; c, distance between the musculo-ligamentous force line and the center of the femoral head; b, the distance between the gravitational force and the center of the femoral head.

and angle α. By choosing arbitrary maximum and minimum values for angle α, most variations in muscle action may be taken into account. Angle α of 50 degrees would correspond to the situation in which abductor action was predominating. Angle α of 10 degrees would indicate predominantly adductor action. As can be seen from the graph, the difference in force on the femoral head, expressed in terms of R_v/G does not vary more than 10 per cent due to differences in angle α.

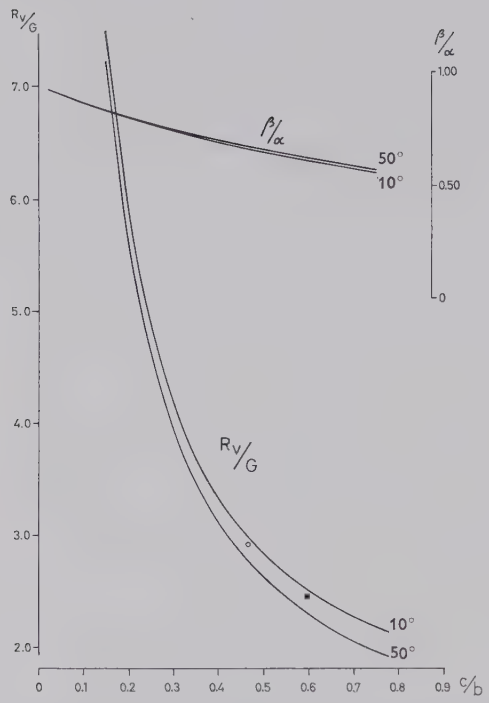


Fig. 2. The relationship of the load on the femoral head to the moments producing pelvic equilibrium. Single leg support and horizontal pelvis. R_v/G, the force on the femoral head in terms of body weight; c/b, the ratio between the musculo-ligamentous force lever arm and the gravitational force lever arm; β/α, the ratio between the angular relationships of the resultant of force line and the musculo-ligamentous force line to the vertical. The light circle (point A), is the value of R_v/G reported by Fischer–Pauwels–Backman; the dark circle (point B), the value of R_v/G reported by Inman. The numbers at the end of the lines refer to angle α.

When the ratio c/b decreases, small changes can produce relatively large increases in R_v/G.

Angle β can be found by substituting in equation 5. β/α is illustrated on the chart as a function of c/b.

DEMONSTRATION OF FORCES ACTING ON THE FEMORAL HEAD

Osborne and Fahrni (1950) have discussed experiments which demonstrated a rise in total pressure on the femoral head when a pelvic specimen was mounted on one femur and loaded with gravitational and simulated muscular forces. Although they gave data for the over-all force on the head, the distribution of

the pressure was not shown, nor was the direction of the reacting force demonstrated experimentally.

It was felt desirable to show that forces simulating muscular and ligamentous pulls could effect the load on the head. A pelvo-femoral preparation was removed during the course of a routine autopsy. All soft tissues were removed except the hip joint capsules and associated ligaments, and the ligaments connecting the bones of the pelvis and lumbar spine. Springs and wires were attached to the pelvis and femurs to simulate the checkrein effects of the short rotator muscles and the ilio-tibial tract. The springs were not under tension until the femurs were adducted more than 10 degrees, or rotated medially more than 20 degrees. The wires were tightened with the femurs in maximum adduction and medial rotation.

A pulley was recessed into the lateral wall of the pelvis near the anterior gluteal line. A hole was drilled transversely through the greater trochanter near its tip, and a steel wire threaded through the hole, over the pulley system, to a weight pan. This wire served as the musculo-ligamentous simulator. A steel spike was placed through the body of the highest remaining lumbar vertebra and weights were placed on this point to simulate G, the body weight (fig. 3 *A, B* and *C*).

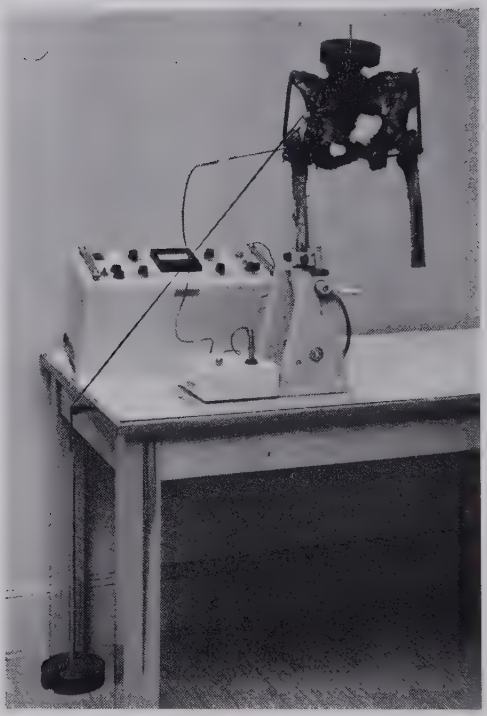
The force acting on the femoral head was measured with a mechanico-electrical transducer. This consisted of three parts. A canula, threaded on the outside; a stylet with a diamond shaped cutting tip; and a stylet containing a longitudinally oriented strain gage (fig. 4). The transducer was connected through a bridge to an Elema Amplifier.

The transducer was sensitive to forces acting in a direction parallel to its longitudinal axis. The position of the transducer in the femur was controlled by means of two-plane roentgenograms. Two positions were used: One was in the superior aspect of the head, with the transducer parallel to the medial trabecular system; the other was in the infero-medial part of the head; the so-called non-weight bearing area (Harrison, Schajowicz, and Trueta 1953). Figs. 5 and 6.

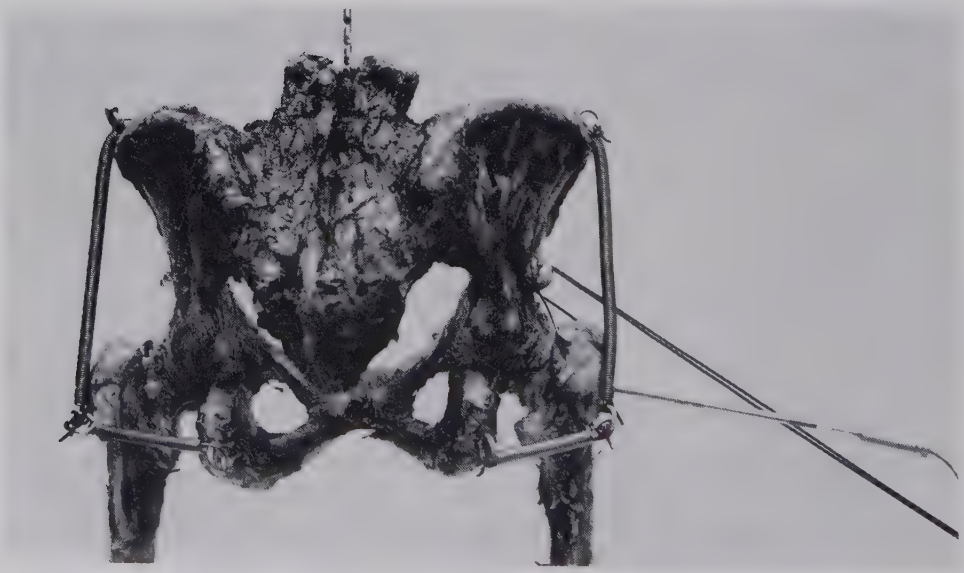
The values read on the amplifier were recorded as 'f-units'. The base line, or zero 'f-unit' level, was recorded by measuring the force acting on the head of the femur in the absence of an added gravitational or simulated muscular force. The change in the force acting on the sensitive tip of the transducer with changes of G and M_v, in reference to the base level, was measured.

Two leg support

Both femoral shafts were held in vises. Measurements were made with increasing values of G, M_v was zero. A progressive increase in the force on the head occurred in position one (parallel to the medial trabecular system). No such increase was noted in position two (infero-medial aspect of head). See fig. 7.



3 *A*



3 *B*

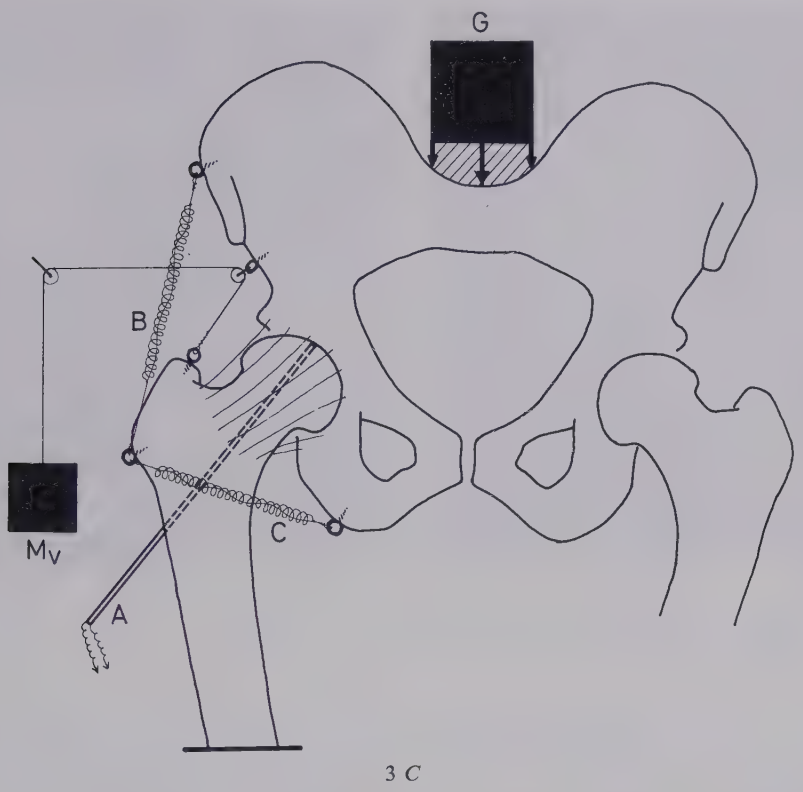


Fig. 3. *A*, pelvic preparation from front. *B*, pelvic preparation from posterior. Springs and wires acting as checkreins. *C*, pelvic preparation (diagrammatic). *G*, gravitational force; M_v, simulated musculo-ligamentous forces; A, transducer; B and C, checkrein springs and wires.

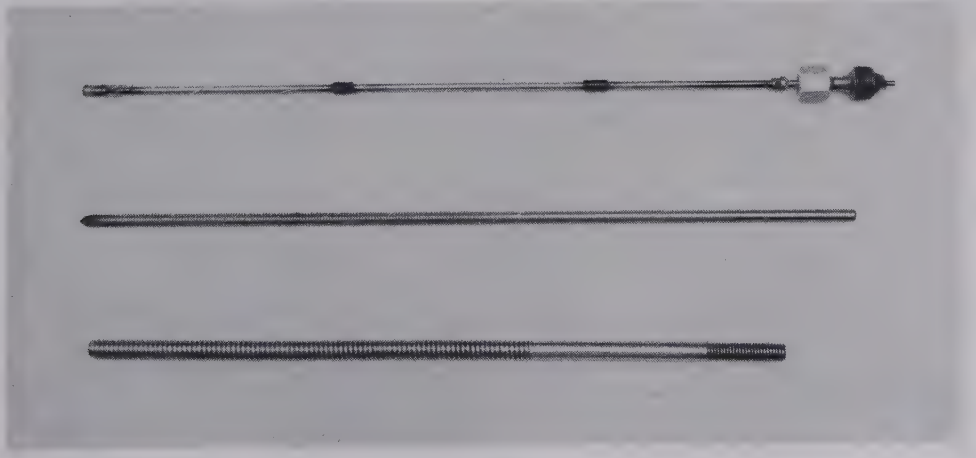


Fig. 4. The mechanico-electrical transducer. (top to bottom) Strain gage stylet, boring stylet, trochar. The trochar is threaded in order to grip the bone.

Fig. 5.

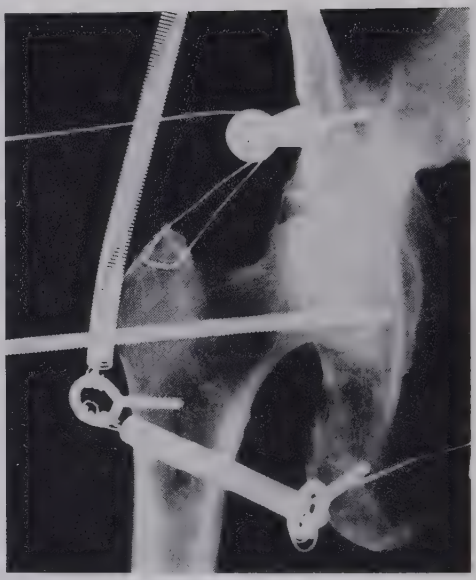


Fig. 6.

Fig. 5. Location of transducer tip in articular cartilage-position one.
Fig. 6. Location of transducer tip in articular cartilage-position two.

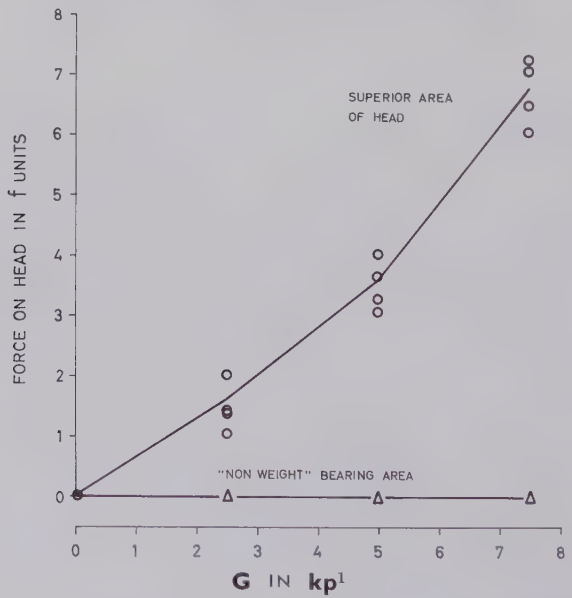


Fig. 7. Changes in force acting on the femoral head with changes in gravitational load. Two leg support. Force is recorded in *f* units which is the amplifier dial reading.

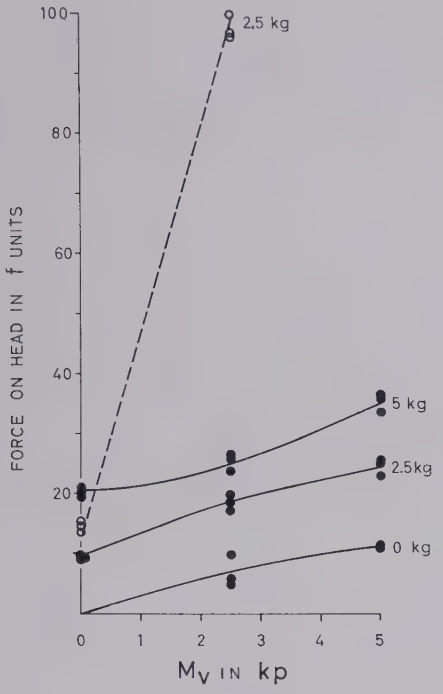


Fig. 8. Changes in force acting on the femoral head with changes in gravitational and simulated musculo-ligamentous forces. Continuous lines, both femurs supported; interrupted line, ipsilateral femur supported. The numbers at the end of the lines indicate gravitational load. The lines indicate values for position one. No increase in force above the base line was found in position two.

Conclusion: The transducer was sensitive to increases in the force on the head. An increase in G of a linear nature produced an increase in the force acting on the superior position of the head. There was no change in the force acting on the femoral head in the so-called non-weight bearing area.

Increasing the value of M_v with constant values for G, produced increasing values for the force acting on the head in position one. No increase was noted at position two (fig. 8).

Conclusion: Simulated forces acting between the tip of the greater trochanter and the lateral pelvic wall produced an increase in the force acting on the superior portion of the head. This has implications when the load on the femoral head is considered in relation to two leg support. It is usually thought that the load on the

[1] The kilopond (Kp) is a measure of force and is defined as the force that gives a mass of 1 kg an acceleration of 9.80665 m/s². The acceleration due to gravity varies between 9.78 and 9.83 m/s². Thus the figure for the force in kiloponds may be regarded as equivalent to the force for the mass in kilograms.

head will be one-half the superincumbant body weight. However, this calculation was made by ignoring possible muscular forces (Backman 1957). With two leg support, the force acting on the femoral head may be greater than that of one-half of the superincumbant weight if muscular and ligamentous forces are present.

Single leg support

One femoral shaft was held in the vise and the muscle simulator loaded, G was constant. Loading M_v produced a greater rise in the force acting on the superior part of the head than when double support was used. Measurements in position two, revealed no increase in force with the loads used during the experiment (fig. 8).

Conclusion: As in double leg standing, loading the muscle simulator produced an increase in the force acting on the superior part of the head. The equilibrium condition was fulfilled by tilting the pelvis.

DISCUSSION

With the understanding that the previously reported studies of the force acting on the femoral head have been based on a similar type of mechanical analogy, data derived from these studies would be expected to fit into the curve displayed in figure 2. Two points are plotted on the curve with data taken from the literature. Point A is from Fischer–Pauwels–Backman (locs. cit.), data based on geometrical analysis and chronophotometry. Point B is from Inman (loc. cit.), data based on geometrical analysis and electromyography. As expected, the two points fall on the curve.

It would seem that the best method available for determining the load on the femoral head would be to study the ratio c/b, as it has the greatest controlling effect on the force. When the ratio is increased it has the effect of lowering the force on the femoral head. A decrease in the ratio brings about the converse.

The values of R_v/G given on the graph in figure 2 are considered to be minimum static loads on the hip during single leg standing. Dynamic components of force will increase the reactive force on the femoral head by up to 50 per cent due to the acceleration and deacceleration of body parts during activity (Pauwels 1935).

This analysis explains how the force on the femoral head increases and decreases depending on the magnitude, direction, and location of the forces involved in pelvic equilibrium.

CHAPTER 3

TRANSMISSION OF FORCE BY THE FEMORAL NECK

It was noted in the previous chapter that the force on the femoral head is considerably greater than that of the superimposed weight itself. This force must be transmitted through the neck of the femur down to the trochanteric area, where it will be partially transmitted back to the pelvis in an internal distribution of force, and partially transmitted through the lower limb to the floor.

The manner in which forces are normally transmitted through the femoral neck will probably have a relationship to the shape and structure of the neck, and the strength of its various parts. Qualitative and quantitative data concerning the manner of action of forces on the upper part of the femur are needed if the mode of transmission of forces through the neck is to be understood.

Calculations for several cross-sections of the femoral neck have been reported by Koch (1917), Grunewald (1920), and Backman (1957) in regard to the state of stress in the femoral neck under various loads and load directions. The latter author, considering the neck as a cortical shell, concluded that the various portions of the cortex were under tension, compression, and shear, depending on the load direction on the head.

The effect of abductor muscle pulls on the state of stress has been demonstrated by Milch (1940) and Pauwels (1948) using the photoelastic method of study. Hirsch and Brodetti (1956 b) have considered the femoral neck to be a tube reinforced by the trabecular systems. Studies were performed using strain gages to show the effect of removing various parts of the trabecular systems on the amount of tension and compression in the cortical shell with vertical loads on the head. Evans *et al.* (1948 a, 1948 b, 1952, 1957) using the Stresscoat method of studying strain have demonstrated the location of strains in the femoral neck subjected to loading. Hirsch and Frankel (1960) have demonstrated that the compressive and tensile strains in the cortical shell of the femoral neck in a pelvic preparation can be altered by forces simulating muscular pulls and loading conditions.

EXPERIMENTAL PROCEDURES

The forces transmitted through the cortical shell of the femoral neck have been studied with the Stresscoat and strain gage techniques. The influence on the pattern of strain distribution produced by alterations in the direction of application of the force acting on the head, and by changes in its magnitude was investigated.

The direction of the force will not only vary with the position of the specimen in space, but also will be varied by the direction and magnitude of the muscular forces demonstrated by the analysis in the previous chapter.

Definition and Measurement of Angles

When discussing directions of force in relation to the anatomy of the upper part of the femur, it is necessary to measure the angle between the shaft and the neck (cervico-diaphysial angle) and the degree of tilt of the shaft from the vertical

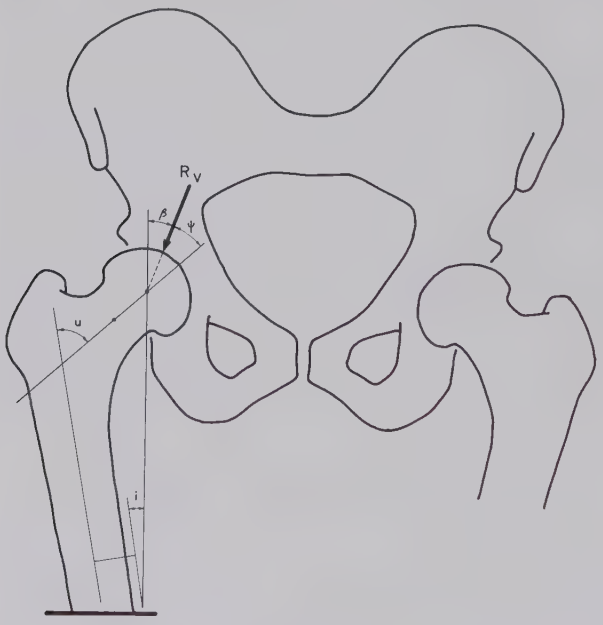


Fig. 9.

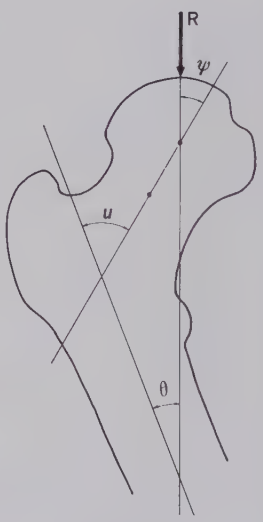


Fig. 10.

Fig. 9. Angular relationships in the upper end of the femur. u, the cervico-diaphysial angle; i, angle of inclination of the femoral shaft during single leg support; β, angle between direction of resultant force (R_v) and the vertical; ψ, angle between the cervical axis and the direction of the resultant of force.

Fig. 10. Angular relationships between forces and axes of proximal end of femur. Θ, angle between force line and diaphysial axis; ψ, angle between force line and cervical axis; u, cervico-diaphysial angle. From Figs. 9 and 10: $\psi \quad u \quad \Theta$; $\Theta \cdot \beta \div i$.

during the test. In figure 9 it is seen that R_v, the resultant force on the femoral head from musculo-ligamentous action and body weight, has a relationship to the cervical axis, angle Ψ. Using a femoral specimen it is possible to reproduce similar angles of Ψ by tilting the femoral shaft by an angle, Θ, and loading the specimen vertically (fig. 10).

The cervico-diaphysial angle (u), the angle between a long axis of the neck and a long axis of the shaft was measured from roentgenograms in the following manner: The specimen was positioned in the vise so that the antetorsion plane of the femur was parallel to the roentgen film. Exposure was then made with a tube distance of 65 cm which was found to be suitable from a technical point of view.

A cervical axis was laid out by drawing a line between two points: one in the center of the head, determined by the concentric circle method of Wiberg (1939); the other point bisected the neck of the femur at its narrowest point (fig. 10).

The axis of the neck in reality is not a straight line but a curved one due to its shape. In this experiment, however, as is commonly done clinically, the axis was drawn as a straight line between consistently measured points, arbitrarily chosen for purposes of geometrical discussions.

Angle u was measured at the intersection of the cervical axis with a longitudinal axis of the shaft, drawn by laying a ruler along the course of the shaft and seeking the best fit. There is a divergence of the diaphysial axis from an ideal axis drawn as a straight line, due to the anterior bowing of the femoral shaft. Although the diaphysial axis had a slightly different curvature from specimen to specimen, the difference was considered to be negligible.

Angle Θ, the angle that the shaft of the femur makes with the vertical force R during the experiment, was determined by the degree of tilt of the vise holding the shaft.

Of importance is the relationship of angle Θ to certain angles present during weight bearing on one leg. These are angle i, the normal inclination of the femoral shaft; and angle β, the angle between the resultant force (R_v) and the vertical (figs. 9 and 10). Values of from 9–15 degrees have been assigned to angle i by Walmsley (1934) and Backman (1957), with the average about 10 degrees.

Angle β has been calculated to be 10–15 degrees by Inman (1947) and 15 degrees by Backman (1957). It can be seen from figures 9 and 10 that $\Theta = \beta + i$. Adding the values of β and i together gives a figure of 19–30 degrees with 20–25 degrees being the most likely range. From this it is concluded that to position the shaft of the femur in the test vise so that the force acting on the head will have a direction near that found on standing on one leg, the shaft must be inclined from the vertical by about 20–30 degrees.

The relation of the line of force passing through the head and neck of the femur

to the cervical axis is expressed by angle Ψ. Backman (1957) calculated angle Ψ to be 28 degrees in single leg stance.

Equation 6 $$\Psi = u - \Theta$$

This can be seen in figure 10.

Stresscoat Studies

Four fresh femoral specimens (all over 50 years of age) were freed from all soft tissue and the surface defatted by means of an ether scrub. Aluminum undercoat was applied with a spray gun, allowed to dry, and the strain indicating lacquer applied in an even coating. The specimens were then allowed to dry from 12 to 24 hours at a constant room temperature of 20°C before the test was performed. The specimen was mounted in the testing machine and the inclination of the shaft to the vertical (angle Θ) noted. The vise rested on two rollers. Force was applied through a well fitting cup. The specimen was loaded and unloaded in 50 Kg increments (fig. 11).

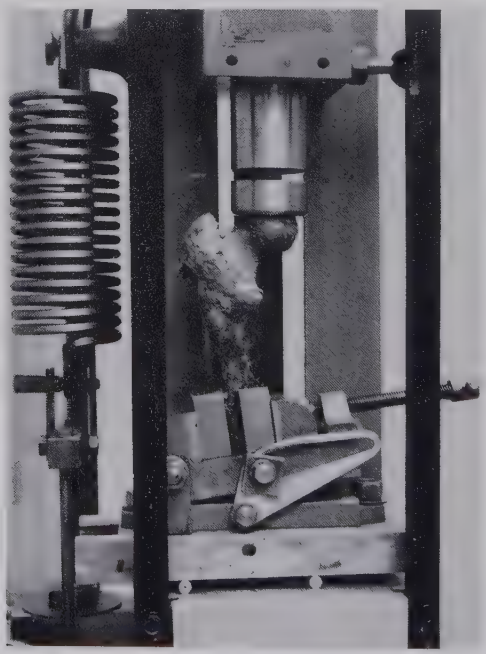


Fig. 11. Arrangement of specimen in compression apparatus during Stresscoat studies. Note rollers under vise, and ball bearing between pressure head and acetabular cup. These measures insured that the force direction was constant.

A *B*

Fig. 12. Stresscoat cracks showing strain pattern in superior cortex of a femoral specimen. *A*, shaft vertical, $\psi = 55°$. *B*, shaft tilted, $\psi = 25°$.

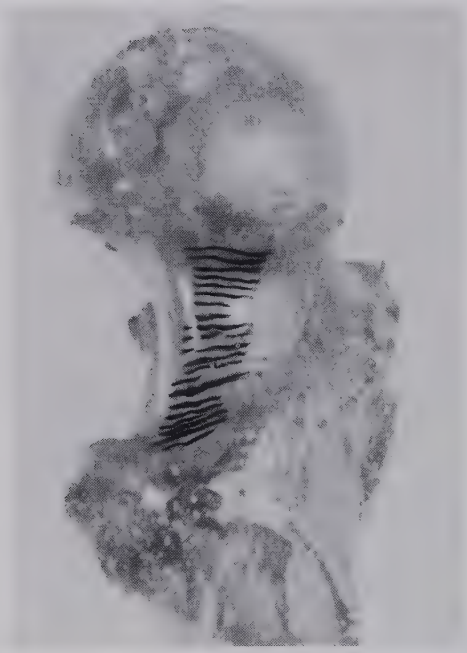
Tensile strain was read during the time the specimen was under compression; compressive strain read after the load was released. The cracks in the lacquer were made more visible for photography by painting them with india ink. As far as possible, the distribution of the initial cracks were noted, along with their progression. This method of study was thought to be useful in a qualitative way, but was not felt to be accurate enough when applied to a bone for quantitative studies, due to variation in the sensitivity of the lacquer.

Results

Shaft vertical ($\Theta = 0$ and therefore $\Psi = u$)

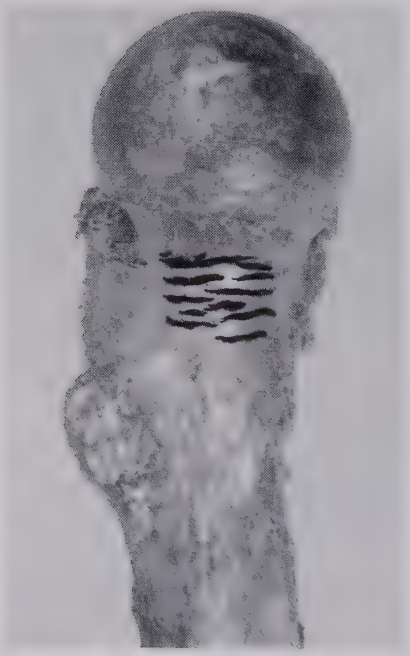
Superior Cortex: cracks were concentrated in an area extending from the midportion of the neck, laterally (fig. 12 *A*).

Inferior Cortex: the cracks were concentrated in an area extending from the subcapital area to the lesser trochanter. There was very little spread around the neck anteriorly or posteriorly with increase in pressure (fig. 13 *A*).



A

C



B

Fig. 13. Stresscoat cracks showing strain pattern in inferior cortex of a femoral specimen. *A*, $\psi = 55°$. *B*, $\psi = 42°$. *C*, $\psi = 25°$.

Fig. 14. Stresscoat cracks showing strain pattern in posterior cortex. *A*, $\psi = 55°$. *B*, $\psi = 42°$. *C*, $\psi = 25°$.

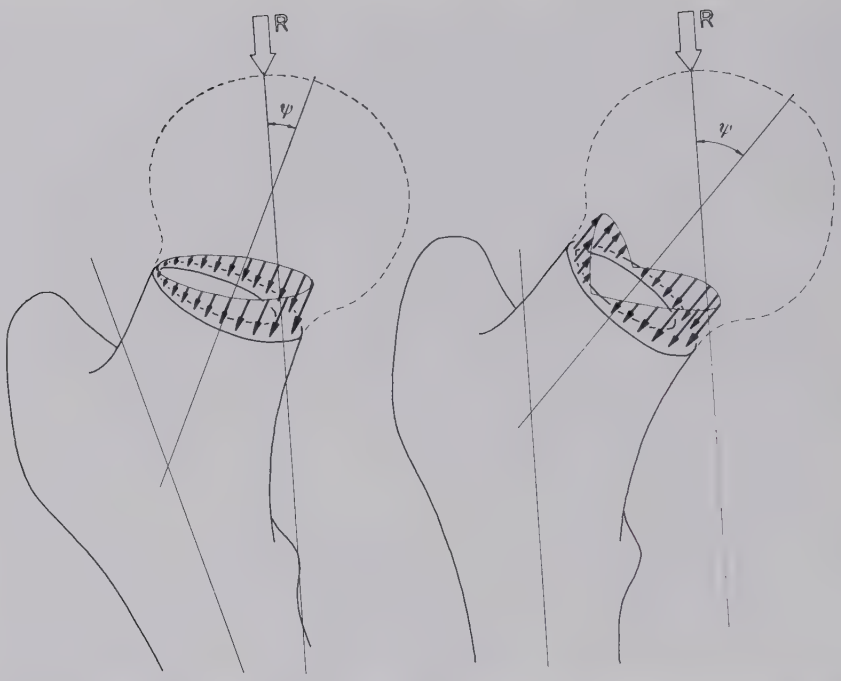


Fig. 15. Distribution of compressive and tensile strains in the cortical shell of the femoral neck (diagrammatic). When the head is loaded with the femoral shaft held vertically (rt), compressive strains are localized in the inferior cortex and tensile strains are found in the superior cortex. There is a steep gradient from compression to tension. (lt) The shaft is tilted, $\psi \approx 25°$. Note difference in strain pattern. The superior cortex is the site of little, if any, strain. The inferior cortex is the site of compressive strains which spread out over the anterior and posterior cortices.

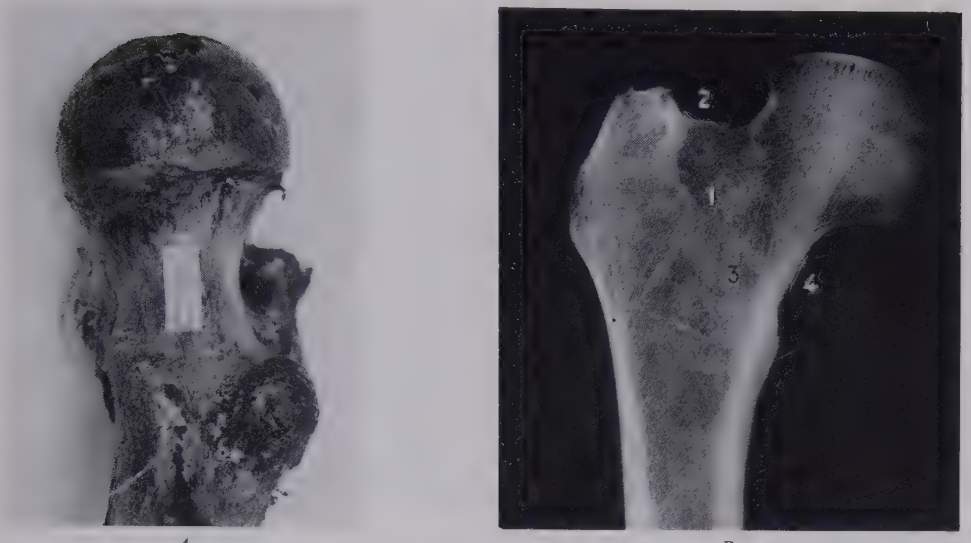
Anterior and Posterior Cortices: a vertical crack was present extending from the superior subcapital area in a direction towards the infero-basal part of the neck (fig. 14 *A*).

Shaft tilted 25–30 degrees (Ψ therefore $= u - 25$ *to 30)*

Superior Cortex: cracks were not always produced, even with the maximum forces used in the test (270–300 Kp.) Cracks that did appear were found due to tension and had a distribution over the middle of the neck (fig. 12 *B*).

Inferior Cortex: the cracks encircled the lower half of the neck close to the subcapital zone (fig. 13 *C*).

Anterior and Posterior Cortices: concentric cracks were produced by compression stresses (fig. 14 *C*).



A *B*

Fig. 16. *A*, photograph of specimen with strain gage mounted on inferior cortex. *B*, roentgenogram of femoral neck with strain gages. 1, anterior; 2, superior; 3, posterior; 4, inferior.

Shaft tilted 14–17 degrees ($\Psi = u - 14$ to 17)

The pattern of cracks showed a transition between the two extremes of $\Theta = 0$ and $\Theta = 25$–30 (figs. 13 *B* and 14 *B*).

Interpretation

The Stresscoat patterns bear out the theoretical calculations that both tensile and compressive strains are present at points on the cortical shell of the femoral neck.

When the shaft is more vertical the strains have a steep gradient from their high concentration of compressive strains in the inferior cortex to the tensile strains in the superior cortex. When the load has a direction considered to be normal during gait, the strain distribution is different. The gradient from compressive to tensile strains around the neck is not as steep, compressive strains occur in the inferior, anterior, and posterior cortices; the superior cortex is the site of slight, if any, tensile strains (fig. 15).

The experiments show that the strain pattern in the cortical shell of the femoral neck varies with the direction of the force acting on the femoral head.

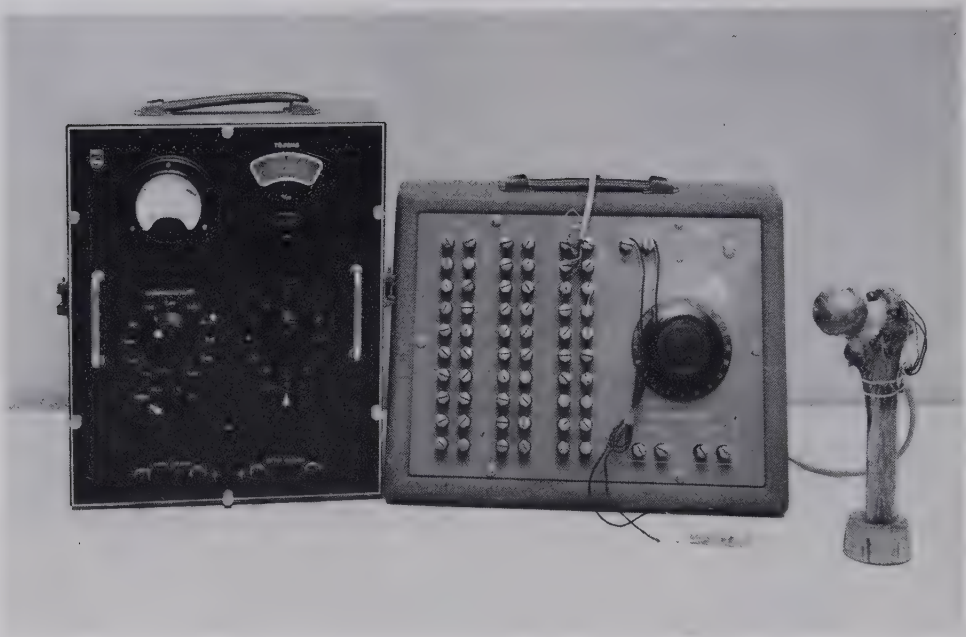


Fig. 17. Strain measuring apparatus. (lt. to rt.) bridge, switch box, femoral specimen with strain gages on neck. (foreground) Strain gage mounted on bone block, used as temperature compensating gage.

Strain Gage Studies

Technique

The same specimens were used as in the Stresscoat studies. Philips Strain Gages ($k = 1.93$, $R - 120.5\ \Omega \pm .5\%$) were glued to the bones using Philips Strain Gage Glue. The technique followed the manufacturer's directions and the experiences of Hirsch and Brodetti (1956) (figs. 16 *A* and 16 *B*).

Readings were made on a Philips battery powered zero reading bridge (fig. 17). The specimens were placed in the test vise, which was mounted on two steel rollers. Force was applied through an acrylic or brass acetabular-shaped cup. A steel ball bearing was interposed between the cup and the compression machine. These steps insured that the direction of applied force was the desired vertical one. It was possible, by means of roentgenograms and photographs to show the relationship of the direction of the force to the structures of the head and neck of the femur (fig. 18). Measurements were made using forces of 50, 100, and 150 Kp, with the shaft of the femur tilted in different positions.

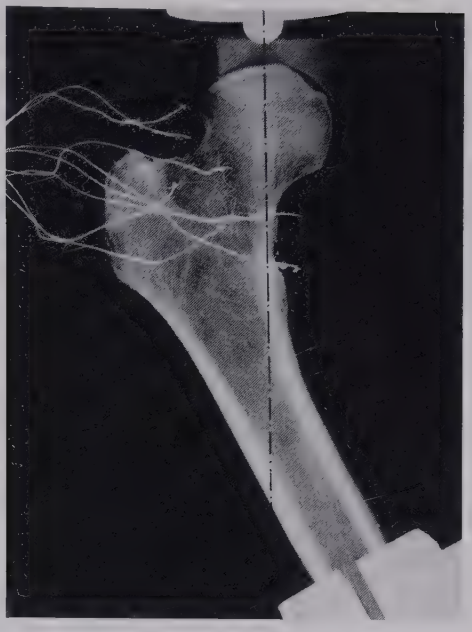


Fig. 18. Roentgenogram of femoral specimen mounted in compression apparatus. Interrupted line shows relation of direction of force to geometry of femoral neck. See Figs. 9 and 10.

Recording of stress and strain

On the charts, strain read from the bridge, and not stress, is plotted. The relationship between stress and strain is given by the equation:

Equation 7 $$\sigma = E \cdot \varepsilon$$

σ is the tensile or compressive stress in the material
ε is the strain
E is the modulus of elasticity.

The modulus of elasticity of bone has been found to vary from 1,800 to 2,400 Kp/mm^2 depending on the type of bone tissue tested (Rauber-Kopsch 1938). It has also been shown to vary within the same general limits in the same bone, depending upon from where the bone specimen has been taken and on its orientation to the axis of the bone (Forssblad 1959). Due to lack of information about the real value of E in the sections used, it is felt to be better to express the findings in terms of the strain.

Results

Figure 19 shows the strains measured at one point on each the inferior and superior cortices, produced with increasing loads and constant shaft angles. It

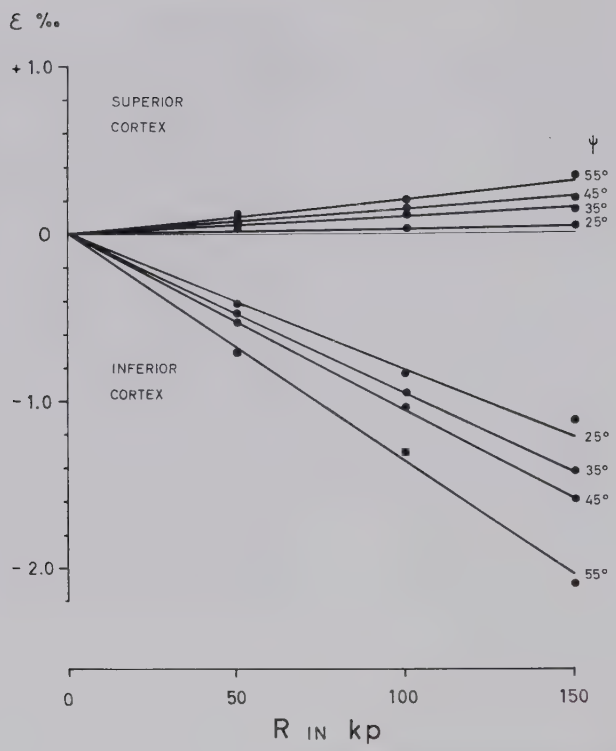


Fig. 19. Stress-strain diagram of femoral neck cortex. The effect on the strain pattern in the cortex of loads on the femoral head (ψ constant).

demonstrates that the relationship is of a linear character within the load limits used, and that the femoral neck follows Hook's Law, confirming the strain gage studies of Hirsch and Brodetti.

Figure 20 shows the effect of varying the direction of the force on the strain at four selected points on the femoral neck. The points are located on the superior, inferior, anterior, and posterior cortices. From the measurements obtained, recorded as strain/load, it can be seen that:

Superior Cortex: there is an increase in the tension strain as Ψ increases, and conversely, a decrease in the strain to near zero levels as Ψ approaches 20–30 degrees.

Inferior Cortex: as Ψ decreases there is a decrease in the compression strain.

Anterior and Posterior Cortices: When Ψ has high values, very little strain of a compression or tension character is found. As Ψ decreases, they are subjected to increasing compression strains.

The measurements made of the strain in the sections of the necks of the femurs

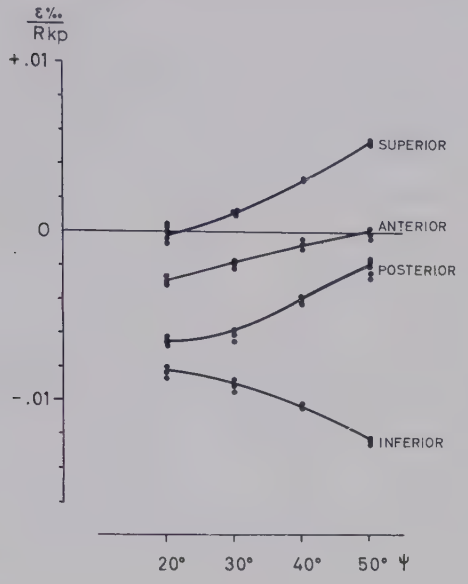


Fig. 20. Stress-strain diagram of femoral neck cortex. The effect of load direction on the strain. Compare with Figs. 12, 13, 14, and 15. When the shaft is vertical, at the location of the strain gages, tensile strain is present in the superior cortex, compressive strain is present in the inferior cortex, some compressive strain is found in the ant. cortex, and little compressive strain is found in the post. cortex. When the shaft is tilted ($\psi \approx 25°$) there is little strain in the superior cortex. The inferior cortex is under less strain than when the shaft is vertical. The anterior and posterior cortices are under more compressive strain. The strain pattern, found when $\psi \approx 25°$, is thought to be the pattern present during single leg stance.

used in these experiments compare well with the theoretical calculations of Backman (1957).

CONCLUSIONS

It was possible to demonstrate that the distribution of forces in the cortical shell of the femoral neck is dependent upon the manner in which force is applied to the femoral head.

The position of the bone and the presence of force vectors from muscle and ligamentous structures are important factors in determining the direction of the resultant force R_r passing through the center of the femoral head and thence through the femoral neck. These two factors will thus have an effect on the distribution of strains in the various parts of the cortical shell.

Stresscoat and strain gage studies have shown that with force directions assumed to be present during locomotion, the strains in the femoral neck are more evenly distributed in a cross-section of the cortical shell than when more vertical directions of force are used.

CHAPTER 4

THE ULTIMATE STRENGTH OF THE FEMORAL NECK

The breaking strength of the femoral neck serves as the limitation to the total force that the femoral neck can transmit. In the past, experiments performed to establish the breaking point of the neck were done with the femoral shaft held vertically or with the bone resting on the greater trochanter. On the whole, the purpose of those studies was to produce an explanation of the fracture mechanism in vivo. No work has previously been reported on the breaking strength with regard to normal angles of load bearing.

VECTORIAL ANALYSIS OF FORCES ACTING ON THE HEAD AND NECK OF THE FEMUR

In the neck, the force R, applied to the head can be divided into two forces acting at right angles to each other. One, the normal force (N), acts along the longitudinal axis of the neck. The other force (P) is a bending force acting at right angles to the long axis of the neck in the frontal plane (fig. 21). They can be found with the following formula:

Equation 8 $$N = R \cos (u - \Theta) = R \cos \psi$$

Equation 9 $$P = R \sin (u - \Theta) = R \sin \psi$$

TEST PROCEDURE

Fresh autopsy specimens were used either immediately or after storage for a few weeks in a deep freezer ($-25°C$). There was no difference in the breaking strength of the bones with either method.

The specimens were mounted in the test vise as previously described. At times a steel ball bearing was interposed between the acetabular cup and the pressure head. It proved difficult to maintain this at forces above 600 Kp due to slipping of

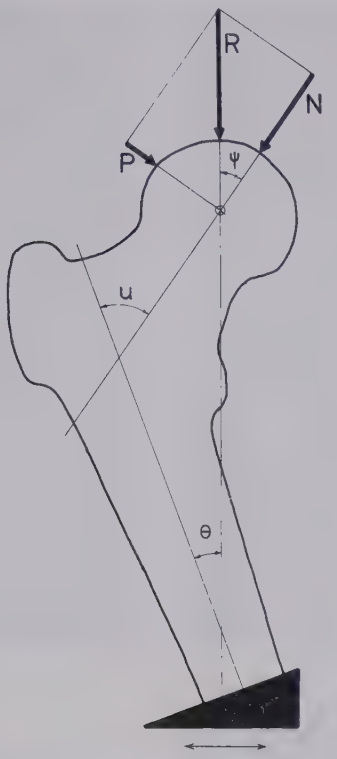


Fig. 21. Analysis of forces acting on the femoral neck. *R*, force acting on the femoral head; *N*, component of force acting parallel to longitudinal cervical axis; *P*, component acting at right angles to cervical axis.

the cup. As this did not seem to make a difference in the results, the bearing was not used routinely. Force was applied by means of the Amsler compression machine at a rate of speed from 50 to 100 Kp/minute until the bone was fractured.

The distribution of specimens according to angle Θ is found in figure 24. It was explained in the previous chapter that values of angle Θ from 19 to 30 degrees would be expected to occur in the subject standing on one leg with the pelvis maintained in a horizontal position. It can be seen from figure 24 that the majority of the specimens were fractured in this range.

CLASSIFICATION OF RESULTS

The fractures were classified into one of three types dependent on the position of the fracture line. The types were subcapital, transcervical, and an intermediary

A

B

C

Fig. 22. Fracture types. *A*, subcapital. *B*, transcervical. *C*, transitional type: subcapital with spike. *D*, photograph of subcapital fracture. *E*, photograph of subcapital with spike fracture. Note small fragment of neck attached to what otherwise would be a subcapital fracture.

D *E*

group which was a subcapital fracture with a spike of neck attached (fig. 22 *A*, *B*, and *C*).

The angle made by the fracture line with the diaphysial axis was measured. As the fracture line was frequently irregular, it was necessary to strike an average. A fracture line parallel to the diaphysial axis was recorded as 90 degrees. A horizontal fracture line, at right angles to the axis of the femoral shaft was recorded as 0 degrees. By doing this, a Pauwels' grade III fracture was registered as 90 degrees if it was parallel to the femoral shaft.

ULTIMATE STRENGTH

The distribution of fractures in relation to breaking force, type and direction, sex and age, normal and bending forces, and the ratio of the normal to the bending force is given in Table 2. Figure 23 gives the distribution of fracture force, sex, and N/P ratios for specimens over 50 years. Specimens with gross pathological changes, as seen on roentgenograms, were discarded.

No good correlation between the breaking force and the type of fracture was found.

Approximately 75 per cent of the male specimens had an ultimate strength of

TABLE 2. *The ultimate strength of the femoral neck.*

(Static force acting in a direction found during one leg stance).

Specimen	Age and Sex	μ	Θ	φ	R (Kp)	N (Kp)	P (Kp)	N/P	Type of Fracture	Fracture Angle.
19 L	75 ♂	57°	20°	37°	970	775	582	1.33	Transcervical	30°
19 R	75 ♂	52°	20°	32°	1010	856	535	1.60	Transcervical	80°
21 L	56 ♂	58°	23°	35°	850	696	488	1.43	Transcervical	110°
21 R	56 ♂	55°	23°	32°	700	560	371	1.51	Transcervical	100°
23 R	77 ♀	57°	23°	34°	450	373	252	1.48	Subcapital with spike	90°
23 L	77 ♀	44°	23°	21°	550	515	193	2.67	Subcapital	45°
24 R	78 ♂	49°	21°	28°	600	531	282	1.88	Subcapital	55°
24 L	78 ♂	51°	21°	30°	625	540	313	1.73	Subcapital with spike	75°
25 L	71 ♀	60°	21°	39°	800	620	504	1.23	Transcervical	105°
25 R	71 ♀	55°	25°	30°	900	770	450	1.71	Subcapital with spike	80°
27 L	70 ♀	41°	21°	20°	365	343	124	2.77	Subcapital	50°
28 R	59 ♀	49°	23°	26°	795	715	350	2.04	Subcapital	70°
28 L	59 ♀	53°	23°	30°	730	630	365	1.73	Subcapital	65°
29 R	83 ♂	59°	15°	44°	680	490	473	1.04	Transcervical	105°
29 L	83 ♂	58°	20°	38°	800	632	492	1.28	Subcapital with spike	50°
30 L	80 ♂	58°	12°	46°	260	180	187	0.96	Transcervical	90°
31 L	60 ♀	46°	25°	21°	500	466	180	2.59	Subcapital	55°
31 R	60 ♀	46°	19°	27°	500	446	227	1.96	Subcapital	50°
32 L	81 ♂	46°	23°	23°	1000	922	390	2.36	Subcapital	50°
32 R	81 ♂	50°	23°	27°	1750	1610	796	2.02	Subcapital	60°
33 L	50 ♀	51°	24°	27°	490	436	223	1.96	Subcapital	50°
33 R	50 ♀	50°	24°	26°	500	450	220	2.05	Subcapital	50°
34 R	86 ♀	49°	19°	30°	660	570	330	1.73	Subcapital	60°
34 L	86 ♀	51°	19°	32°	740	592	392	1.51	Transcervical	85°
35 R	42 ♀	48°	24°	24°	770	705	313	2.25	Subcapital	45°
35 L	42 ♀	46°	17°	29°	720	630	349	1.81	Subcapital	40°
36 L	83 ♀	53°	25°	28°	610	540	287	1.88	Subcapital	50°
36 R	83 ♀	50°	15°	35°	470	385	270	1.43	Subcapital with spike	55°
37 L	77 ♀	46°	14°	32°	750	635	398	1.60	Transcervical	90°
37 R	77 ♀	42°	14°	28°	1000	884	470	1.88	Subcapital with spike	90°
39 R	80 ♀	53°	11°	42°	440	327	295	1.11	Transcervical	90°
39 L	80 ♀	42°	25°	19°	340	320	111	2.88	Subcapital	25°
40 R	70 ♂	47°	9°	38°	975	770	600	1.28	Transcervical	95°
40 L	70 ♂	48°	9°	39°	980	760	617	1.23	Transcervical	65°
41 R	68 ♀	54°	22°	32°	740	629	392	1.60	Transcervical	90°
41 L	68 ♀	55°	26°	29°	680	595	330	1.80	Subcapital with spike	65°
42 L	75 ♀	50°	23°	27°	620	553	282	1.96	Subcapital	60°
42 R	75 ♀	45°	11°	34°	470	390	263	1.48	Transcervical	90°
43 L	82 ♀	50°	22°	28°	590	522	277	1.88	Subcapital	45°
43 R	82 ♀	49°	22°	27°	480	428	218	1.96	Subcapital	50°
44 L	72 ♂	50°	21°	29°	1000	875	485	1.80	Subcapital	90°
44 R	72 ♂	50°	21°	29°	1060	928	514	1.81	Subcapital	75°
45 R	62 ♂	59°	21°	38°	1430	1127	879	1.28	Transcervical	70°
45 R	62 ♂	56°	22°	34°	1420	1179	795	1.48	Transcervical	80°
46 R	71 ♀	51°	20°	31°	680	585	350	1.67	Subcapital with spike	70°
46 L	71 ♀	53°	18°	35°	850	697	488	1.43	Transcervical	90°

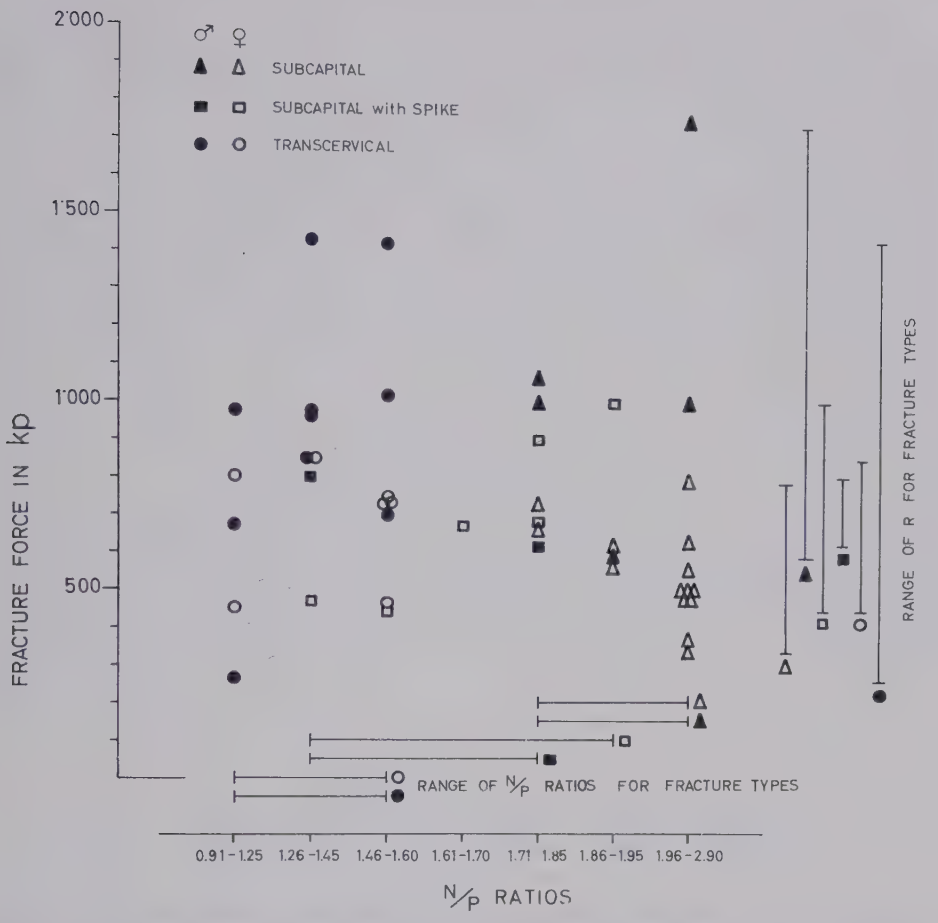


Fig. 23. The ultimate strength of the femoral neck. Distribution of specimens by sex, fracture force, fracture type, and ratios of normal/bending components of force. There is no correlation between type of fracture and breaking force. Breaking forces (R) tend to be higher in males. There is a correlation between type of fracture and N/P ratio. Transcervical fractures occur with ratios below 1.60. Subcapital fractures occur with ratios above 1.70. The subcapital with spike fractures occur in the middle range of ratios. All specimens over fifty years of age.

800 Kp or more, while nearly 90 per cent of the female specimens broke at 800 Kp or less. The mean for men was 971 Kp (260–1750); for women the mean was 617 Kp (340–1000). The sex difference was thought to be due to the presence of more bone tissue in the masculine specimens.

The ratio N/P is derived from angle Ψ. Angle Ψ is equal to the cotangent of the ratio N/P. There is no good correlation between the fracture force and angle Ψ within the limits used in the experiment.

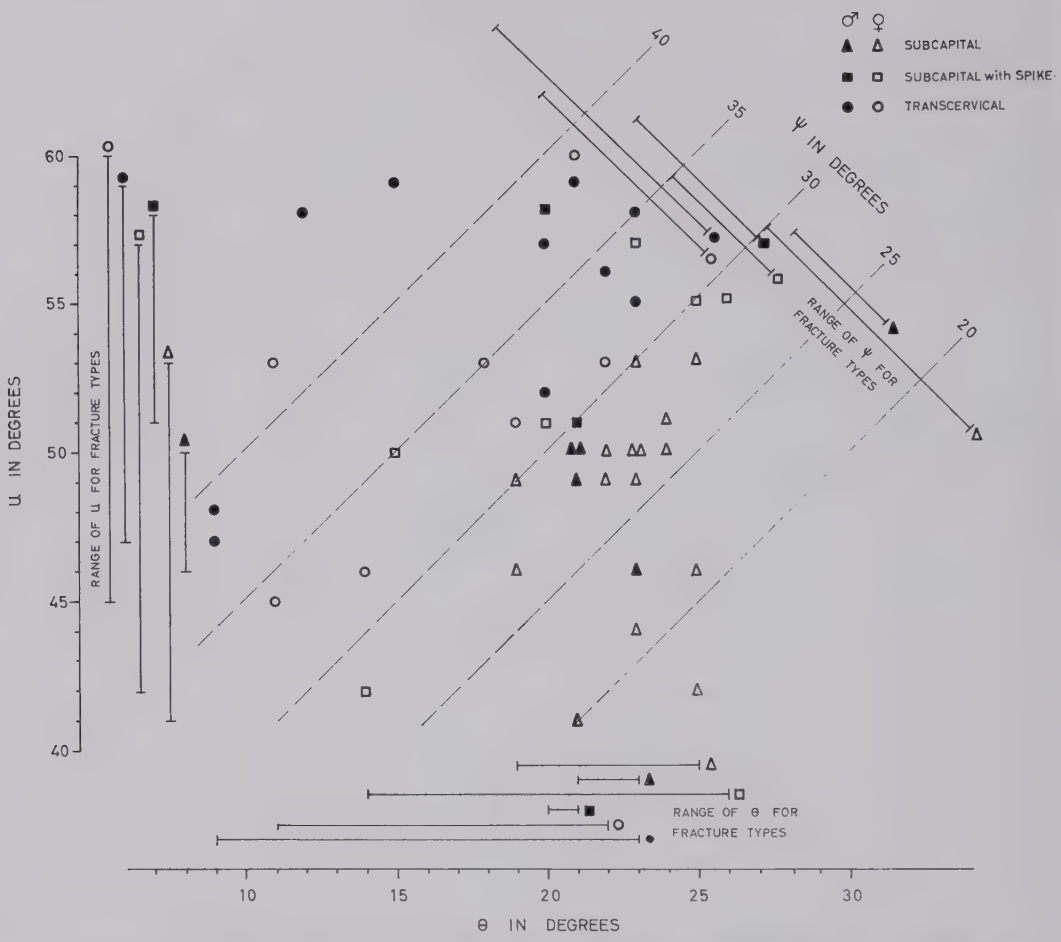


Fig. 24. The ultimate strength of the femoral neck. Distribution of specimens by sex, angles u, Θ, and ψ, and fracture type. Most of the specimens were positioned so that angle Θ had values consistent with the force line during single leg stance (19–30°). There is no strong correlation of fracture type with either u or Θ. There is a positive correlation between fracture type and angle ψ. All specimens over fifty years of age.

CORRELATION OF FRACTURE TYPE WITH FORCES IN THE FEMORAL NECK

Figure 24 gives the distribution of values for angles u, Θ, and Ψ, for the type of fracture, and for the sex of the specimen. The type of fracture did not seem to be determined to a great extent by either angle u or angle Θ. There was a positive correlation of fracture type to angle Ψ.

Correlation of the N/P ratio with the fracture type is seen in figure 23. All subcapital fractures occurred above a ratio of 1.70, corresponding to $\Psi = 30.5$ degrees. The transcervical fractures occurred with ratios below 1.60, equivalent to Ψ equal to 32 degrees. The subcapital fractures with a cervical spike occurred in the middle range of N/P values.

Table 1 contains information pertaining to fracture experiments reported in the literature. Most of the specimens were broken with the shaft held vertically or nearly so. The values for angle Ψ would be expected to be high, close to the values for angle u. The ratio of N/P would then be low, and the fractures would be expected to fall into the transcervical group, which in fact they do.

CORRELATION OF FRACTURE TYPE WITH THE STATE OF STRESS IN THE FEMORAL NECK

From the Stresscoat and strain gage studies performed on the femoral neck with different values of angle Ψ, it was seen that there was a different pattern of stress distribution depending on the value of this angle (fig. 15). By examining the fractured specimens with this strain pattern in mind an explanation for the various sites of failure was made.

Subcapital fractures

These were found where angle Ψ was less than 30.5 degrees. Under these circumstances the superior cortex was under slight tensile strain, slight compressive strain, or strainless. Compressive strains were distributed in the anterior, posterior, and inferior cortices, with a concentration near the subcapital area.

During the fracture experiment the bone broke slowly as a rule, and it was possible to take roentgenograms while the fracture process was taking place. The bone seemed to fail by compression in the inferior subcapital area; however, if the value for angle Ψ was very low, the bone failed by compression in the superior cortex. This seems to fit in with the finding of compressive strains in these areas.

At times part of the subcapital area of the bone was found to have failed while others areas were still intact. Figure 25 *A* shows failure in the superior cortex while the inferior cortex and the medial trabecular system remained intact. Figure 25 *B* demonstrates failure of the superior cortex, the inferior cortex, and part of the medial trabecular system. Figure 26 shows a complete fracture without displacement, and figures 27 *A* and 27 *B* show a displaced fracture. Specimens were also found to displace into a valgus position.

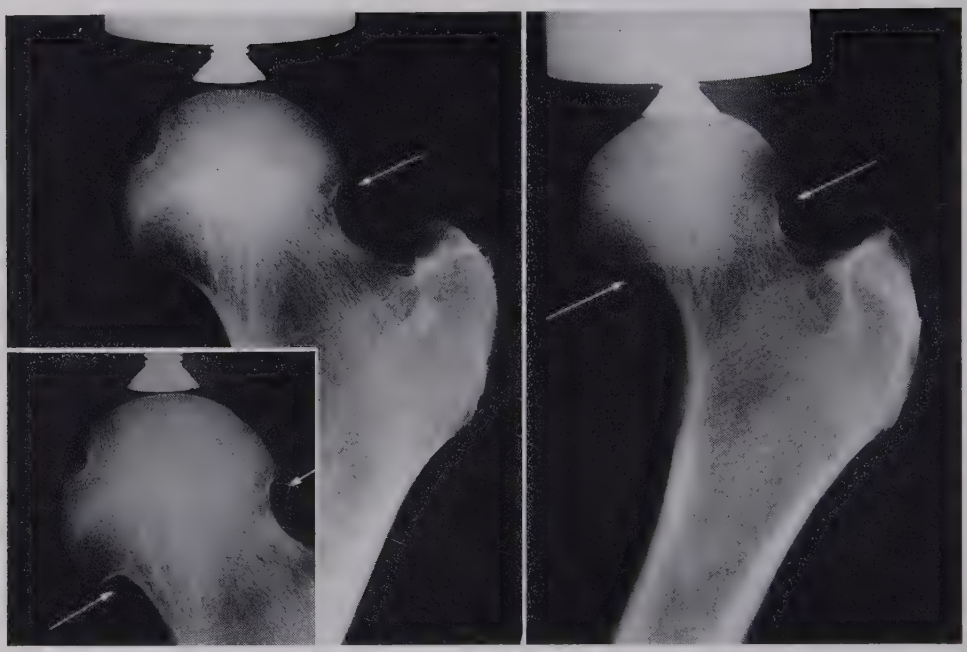


Fig. 25 *B* Fig. 25 *A* Fig. 26

Fig. 25. Stages in production of a subcapital fracture. *A*, disruption of the superior cortex. The inferior cortex and the medial trabecular system are intact. *B*, (inset) the fracture has involved the inferior cortex and part of the medial trabecular system.

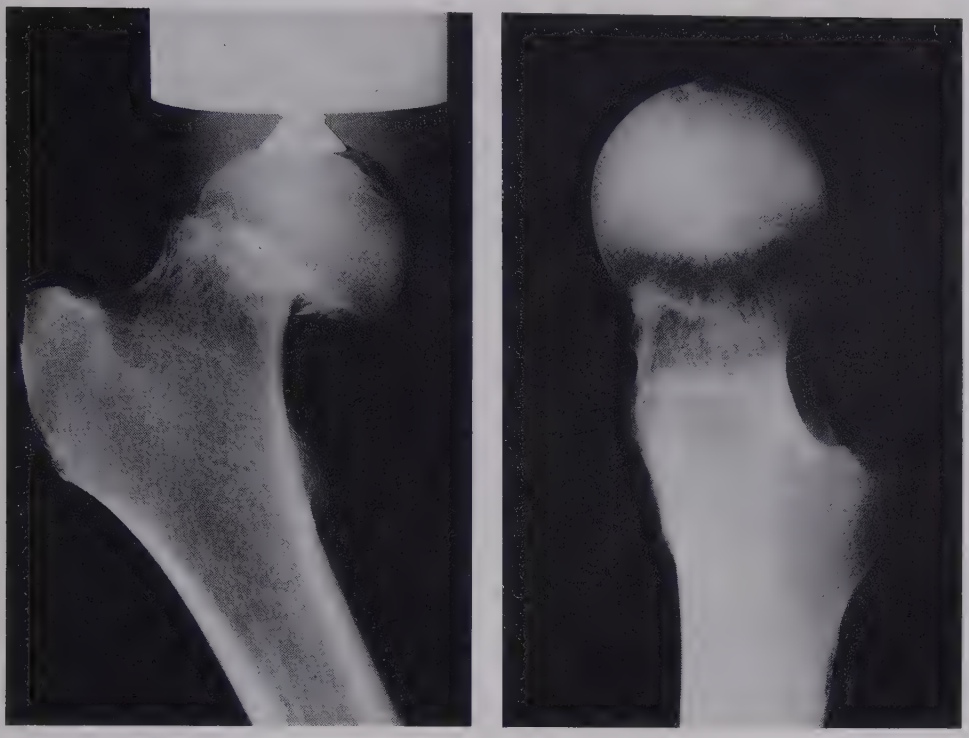
Fig. 26. Completed subcapital fracture without displacement.

Transcervical fractures

These fractures occurred with higher values for angle Ψ and smaller N/P ratios. The bending moment on the neck was higher than with subcapital fractures, with a steeper gradient of strain distribution.

The bones usually fractured suddenly. They failed by rupture of the superior cortex due to tensile strains. At times failure progressed in an observable manner, with the inferior cortex failing only after the force acting on the head was increased.

Rupture of the superior cortex occurred either at the subcapital zone or at a point lateral to this. Failure of the inferior cortex usually occurred at the junction of the medial trabecular system with the cortical shell, although it could be seen to break either proximally or distally to this region.



A *B*

Fig. 27. Displaced subcapital fracture. *A*, anterior-posterior projection. *B*, lateral projection.

Intermediary Type

These fracture lines were reminiscent of subcapital fractures, but differed by always having a fragment of neck attached to the head of the femur. These fractures occurred in the middle range of values for angle Ψ, where the cortical shell had a transitional distribution of strain.

IMPLICATIONS OF THE STUDY TO THE FRACTURE MECHANISM *IN VIVO*

Despite the limitation of this study to one plane, the frontal, it would seem that it offers an explanation for the production of subcapital and transcervical fractures during life. It has been shown by Hirsch and Frankel (1960) that by applying a compression to the longitudinal axis of the neck, the type of fracture produced

by vertically loading the femoral head could be influenced. This compression had the effect of simulating muscular forces. Other experiments have only to a slight extent, if at all, simulated muscular forces acting over the hip joint. Backman (1957) disregarded the effect of muscular contraction in his studies. Although he was able to produce medial fractures by loading the trochanter, he was unable to correlate the fracture type with the stresses involved, and found it impossible to predetermine the type of fracture produced. Backman noted that the type and direction of force were of no significance in producing subcapital fractures.

In this study, however, it was possible to produce subcapital and transcervical fractures in a consistent manner by varying the resultant of the forces acting on the femoral head and neck. By doing this, the ratio of bending and normal components of force was altered, with consequent change in the strain distribution in the cortical shell.

It was possible to predetermine the fracture type in many of the experiments.

The type of fracture was only influenced by the direction of the force when static loads were used.

CHAPTER 5

PRELIMINARY STUDIES OF INTERNAL FIXATION

The premise upon which all currently used fixation material is based, is that the fracture can be immobilized by connecting the head of the femur to the neck by means of an appliance anchored to the femoral shaft. A great variety of methods have been used. Basically, however, the screw, nail, threaded wire, or other ap-

TABLE 3. *The effects of load on osteosyntheses.*

(Static force parallel to the femoral shaft).

Specimen	Age and Sex	φ Before test	φ After test	Breakdown force (Kp)	Observations
Charnley Appliance					
1 R	75 ♀	120°	117°	100	Inferior cortex of neck fractured
2 R	78 ♀	120°	120°	110	Inferior cortex of neck fractured
3 R	74 ♀	120°	117°	245	Inferior cortex of neck fractured
4 R	77 ♂	120°	—	190	Inferior cortex of neck fractured
5 R	68 ♀	120°	117°	120	Head displaced 0.5 cm on neck
6 R	70 ♀	120°	116°	260	Inferior cortex of neck fractured
7 R	74 ♀	120°	—	175	Inferior cortex of neck fractured
8 R	77 ♂	120°	—	175	Inferior cortex of neck fractured
9 R	69 ♂	120°	—	140	Inferior cortex of neck fractured
McLaughlin Appliance					
2 L	78 ♀	145°	130°	290	0.5 cm head displacement, inf. cortex fractured at 200 Kp
3 L	74 ♀	130°	125°	290	Inferior cortex of neck fractured
4 L	77 ♂	130°	125°	360	Inferior cortex of neck fractured
5 L	68 ♀	135°	133°	205	Inferior cortex of neck fractured
7 L	74 ♀	120°	115°	165	Slight inferior cortical fracture on nail insertion, increased at breakdown force
8 L	77 ♂	120°	115°	230	
9 L	69 ♂	115°	112°	120	

Fig. 28 *A*

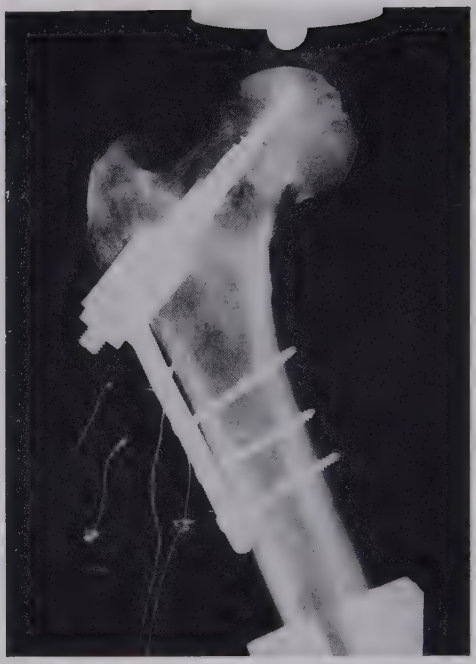


Fig. 28 *B*

pliance has a firm hold on the head, and a more or less firm hold on the lateral femoral shaft. From this standpoint, there is no basic difference between nails and screws and those constructions which have a side bar or pin, except in the firmness of attachment of the appliance to the femoral shaft.

Of the forces which must be transported from the head of the femur to the shaft, part can be carried by the bone, especially if the fracture is of a stable type, e.g. valgus or Pauwels' grade 1. Part of the load must be transmitted by the appliance. The efficiency of immobilization thus depends upon several factors, the inherent stability of the fracture, the strength of the appliance, and the strength of the bone.

The aim of the preliminary studies was to determine how an osteosynthesis could be analyzed experimentally. It was believed to be of advantage in analyzing the strength of the construction to study various parts of the upper end of the femur in regard to the osteosynthetic material.

Determination of the load carrying ability of an osteosynthesis was performed after a femoral neck was divided by sawing perpendicular to its long axis and reunited with an appliance. Tests were carried out with the shaft held vertically in an immovable vise.

C

Fig. 28 *C* and *D*

Fig. 28. Specimen tested with Charnley compression screw-plate. *A*, before test; *B*, after test. Downward bending of the appliance has occurred. *C*, photograph of neck showing fracture of inferior cortex. *D*, roentgenogram of cross-sections of head and neck. There is ovalization of the screw hole in the neck and a fracture of the inferior cortex. The outline of the appliance in the head is not distorted. Reproduced through the courtesy of the Editor of Acta chirurgica Scandinavica.

Observations were made concerning the breakdown force, the plastic deformation of the appliance, and the resistance of the bone to crushing. Data are presented in Table 3.

Failure was almost universally due to a fracture of the inferior cortex of the femoral neck. This fracture was produced through the mechanism of the appliance bending under load, cutting the cancellous bone of the neck, resting on the cortical shell, and finally breaking this structure (fig. 28). The femoral head was not found to be crushed in any test. These findings were confirmed by sectioning the femoral specimens and studying the trabecular pattern in the head and neck (fig. 29).

To demonstrate the resistance of the head to crushing, excised heads were mounted on nails and subjected to compression. Two observations were made:

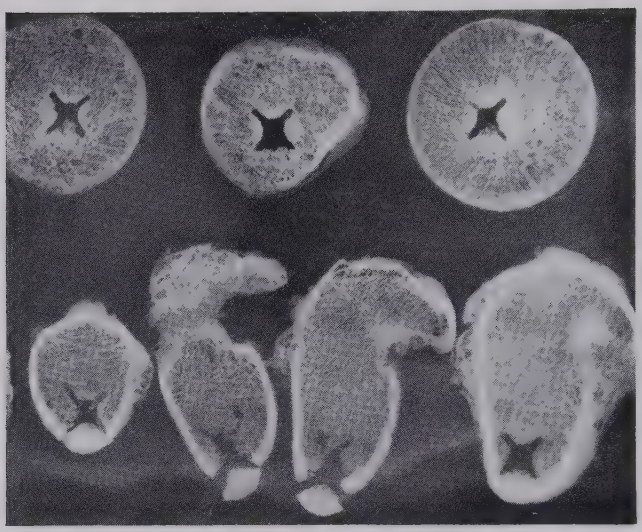
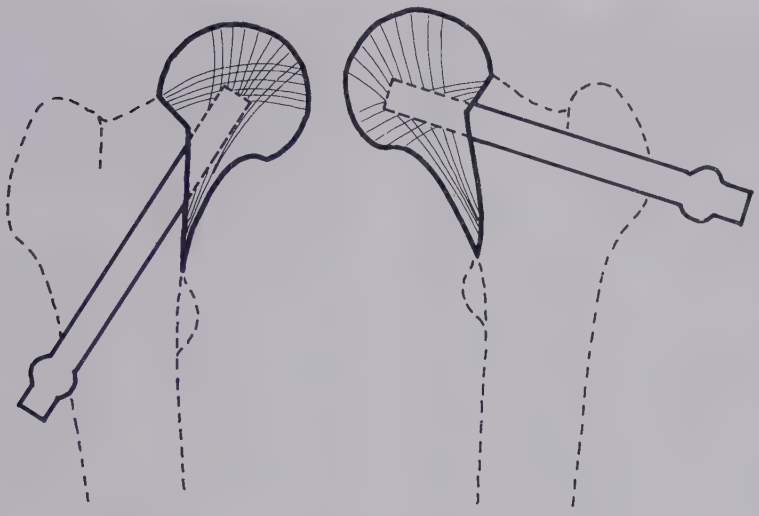


Fig. 29. Roentgenogram of cross-sections of the femoral neck and head of a specimen tested with the McLaughlin nail-plate. The nail bent under load and produced a fracture of the inferior cortex. The outline of the four flanged nail is sharply defined in the head, showing that no crushing has taken place. Reproduced through the courtesy of the Editor of Acta Chirurgica Scandinavica.

One, the ability of the head to resist crushing and cutting through by the nail depended on the position of the nail in the head, a horizontal nail being superior to an obliquely directed nail. Secondly, bending of the nails occurred before they could crush the head in many cases. Data from the tests are found in Table 4; experiments are illustrated in figure 30.

TABLE 4. *The resistance of the femoral head to crushing.*

Specimen	Age and Sex	Position of Nail	Crushing Force (Kp)	Comment
10 L	78 ♀	Horizontal	400	Nail broke at 400 Kp
11 R	65 ♂	Horizontal	400	
11 L	65 ♂	Steep	200	
12 R	70 ♀	Horizontal	210	Nail slipped in vise, two loadings
12 L	70 ♀	Steep	90	
13 R	66 ♀	Horizontal	180	Nail bent at 150 Kp
13 L	66 ♀	Steep	105	
14 R	68 ♂	Horizontal	300	Nail bent at 240 Kp
14 L	68 ♂	Steep	200	
15 R	63 ♂	Horizontal	240	Nail broke at 240 Kp
15 L	63 ♂	Steep	125	



A *B*

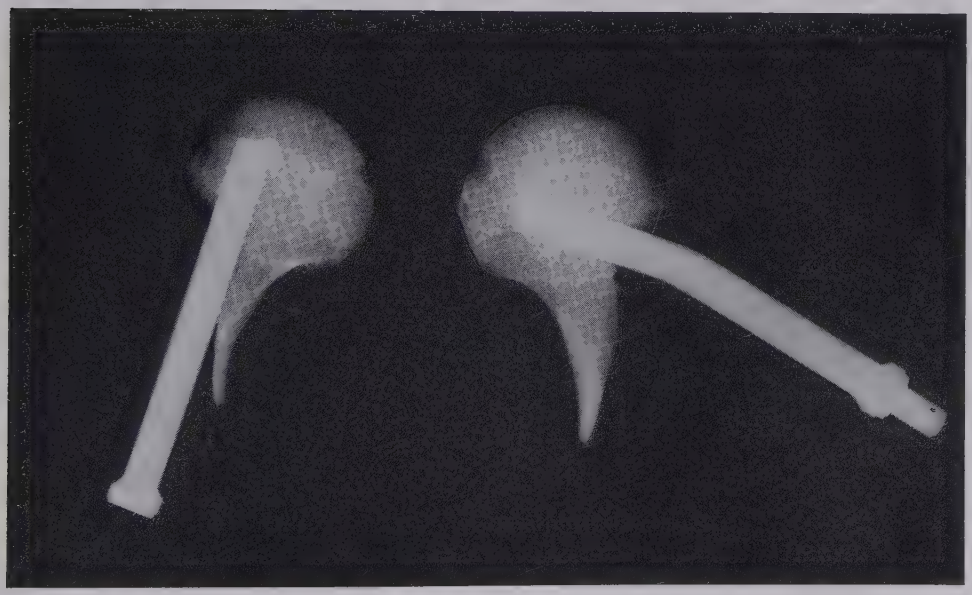
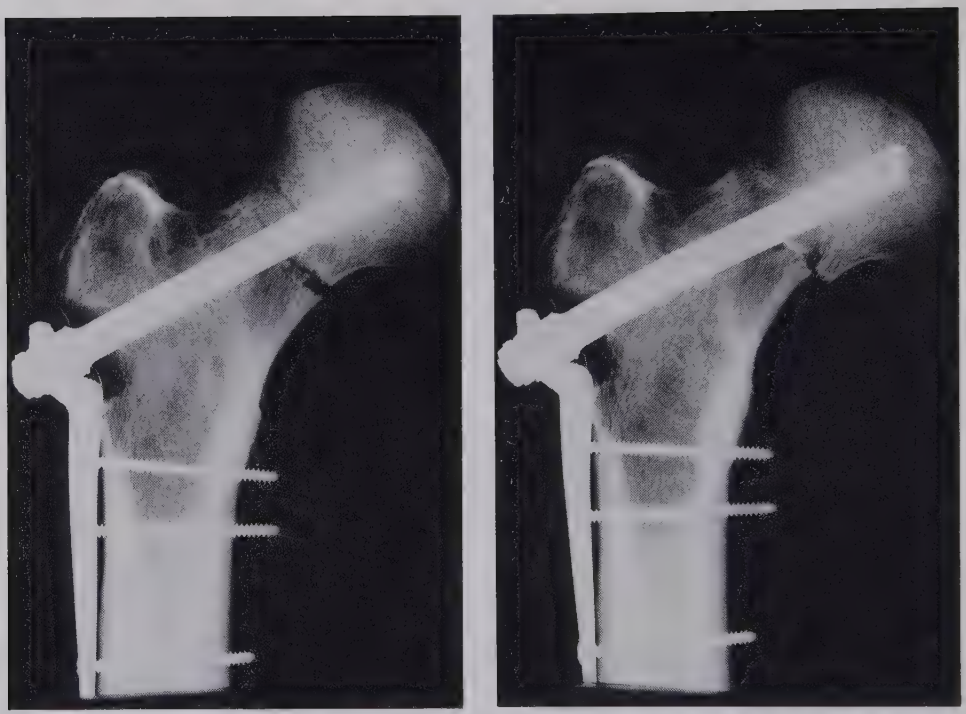


Fig. 30. Diagrams showing orientation of the femoral head and the nail during tests on crush resistant strength of the head. *A*, horizontal nail which has bent under load before the capital bone was crushed. *B*, nail in steep position which has migrated inwards. Reproduced through the courtesy of the Editor of Acta Orthopaedica Scandinavica.



A *B*

Fig. 31. Serial roentgenograms of specimen tested with McLaughlin appliance. Shaft tilted ($\psi = 25°$) *A*, before test. *B*, after loading. The nail has deformed.

The femoral cortex holding the plate screws was strong as far as screw holding ability was concerned; no plate pulled away because of failure in this region.

These tests had been performed with the shaft held vertically. Thus the reacting force on the femoral head was not at a normal angle as far as standing on one leg was concerned, although this load direction might have a relation to the action line during other states.

The tests were repeated using force reaction directions consistent with a one leg stance to see if the same phenomena were present. The specimens were not loaded to failure, but the tests stopped when the load reached 200–300 Kp.

Through the use of serial roentgenograms, made while the specimens were being loaded, it was possible to study the downward bending of the appliances (figs. 31 and 32). Data for the specimens used with McLaughlin nail-plates and Charnley compression screw-plates are found in Table 9.

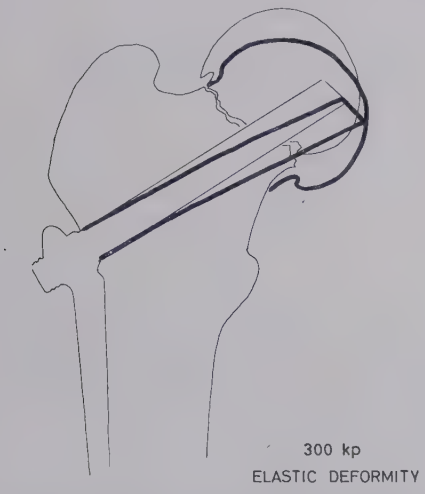


Fig. 32. Superimposed tracings from serial roentgenograms of specimen tested with a McLaughlin nail-plate. Thin line, before test. Dark line, during test. An elastic deformity of the appliance is present.

When the femurs were sectioned, the cancellous tissue of the neck was found to be crushed, the head remaining intact (fig. 32).

From these experiments and observations a hypothesis was drawn that there were relatively weak and strong areas in an osteosynthesis. The head and shaft cortex below the greater trochanter were strong, and the bony tissue of the neck and the appliance were parts where failure occurred. To have the strongest possible osteosynthesis, a major part of the force acting on the head of the femur should be transmitted to the femoral shaft via the nail and plate.

The question then arose: How strong could an osteosynthesis be made using internal fixation? In finding the answer, it was necessary to study the appliances in more detail; and in particular to investigate the theory behind them. The possibilities of making stronger appliances were investigated; methods of testing the appliances both out of the bone and *in situ* had to be developed.

CHAPTER 6

APPLIANCES

STATIC STRENGTH OF APPLIANCES

It is desirable to protect the tissue on which the nail or screw rests in the femoral neck, by having minimal elastic deformations and insignificant plastic deformations when load is applied to the femoral head. These deformations will depend on the design of the appliance and the material used. Deformations can be measured by means of loading and unloading the appliance, and noting the deflection at a known distance from the tip of the nail or screw. The load test can be plotted in a diagram showing the elastic and plastic deformations for given loads (fig. 33).

The yield stress, σ_y, is that stress at which increasing plastic deformations are produced with increasing load. Below this limit, the appliance acts like an elastic body; that is, there is a return to the original dimensions after a deforming load is removed.

The necessary minimum yield strength can be calculated from the formula:

Equation 10

$$\sigma_y = \frac{P_n L}{I} \cdot \frac{h}{2} \qquad \text{(Symmetric nail section)}$$

P_n is the expected load on the nail, perpendicular to the nail
I is the moment of inertia of the nail section
L is the length of the nail
h is the vertical height of a cross section of the nail.

In figure 34, the minimum yield strength required (σ_y), and the moment of inertia of the nail section for different shapes of the nail cross section are presented.

An Idealized Appliance

Certain characteristics of nail-plates can best be understood if a stylized design is utilized and the mechanical behavior of appliances deduced by means of appropriate calculations. A design for such an appliance is shown in figure 35.

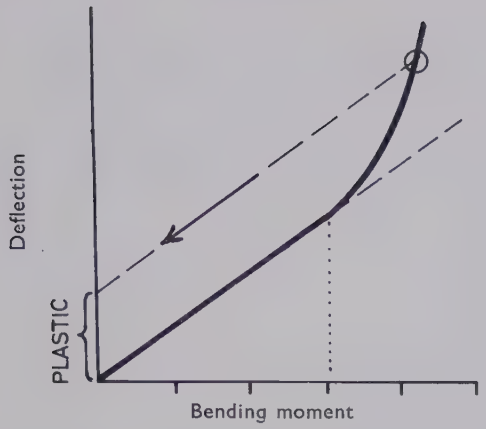


Fig. 33. Load-deflection diagram. Bending moment (Kpmm) is plotted against deflection (mm) at tip of nail. Dotted line shows the point at which yield begins. When unloading takes place from bending moments above the yield limit the appliance no longer returns to its original dimensions but has a permanent (plastic) deformity.

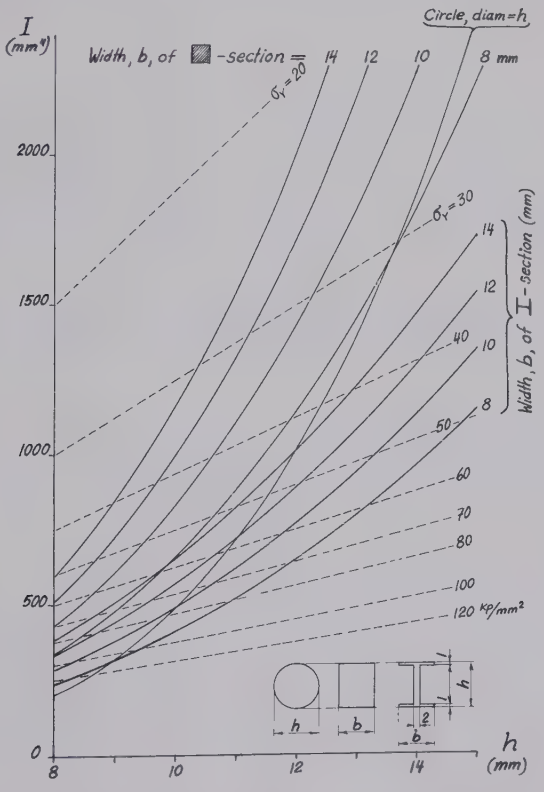


Fig. 34. The yield strength (σ_y) of material required for appliances of various shapes, no yielding to take place below 7500 Kpmm. I, the moment of inertia; h, height of the nail cross-section; b width of the nail cross-section. Three examples of cross-section shape are given: square, circle, 'I' beam.

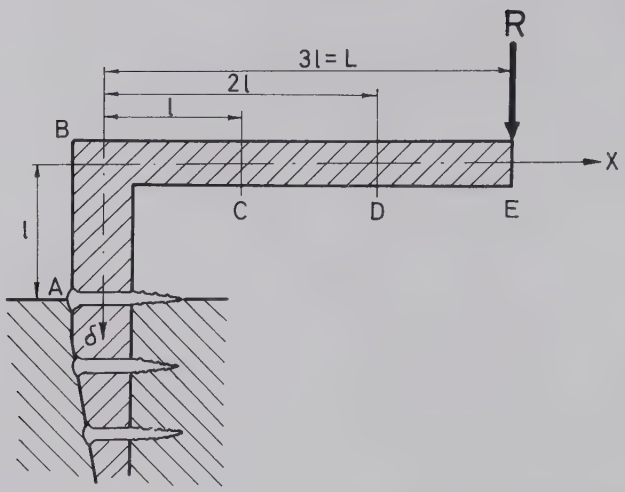


Fig. 35. An idealized appliance. x, the central axis of the nail; l, the distance between the nail-plate junction (B) and the top screw of the plate (A). Points C, D, and E are found along the nail at distances l, 2 l, and 3 l from B.

It is of interest to know the deflections at various points along the course of the nail, and also the effect on this deflection of varying the stiffness of the nail and the plate.

The deflection, D_x, at a specified point, X, on the nail can be found by means of the following equation:

Equation 11

$$D_x = \frac{P L^3}{E I_{be}} \left[\frac{1}{3} \frac{I_{be}}{I_{ab}} \frac{X}{L} + \frac{X^2}{L^2} - \frac{1}{3} \frac{X^3}{L^3} \right] = \frac{P L^3}{E \cdot I_{be}} \cdot d_x$$

E is the modulus of elasticity
I_{ab} is the moment of inertia along AB
I_{be} is the moment of inertia along Be (can be found from figure 34)
d_x is the dimensionless factor in the parenthesis [ ].

The influence of the stiffness of the plate

The importance of the stiffness of the plate to the deflection of the nail can be calculated from equation 11. A few examples are in table 5:
It can be seen that the plate design and material play a large role in the deflections of the nail.

TABLE 5.

	Deflection factor, d_x at point		
	C	*D*	*E*
If $I_{ab} = \frac{1}{2} I_{be}$ (pliant plate)	.333	.888	1.333
If $I_{ab} = I_{be}$	.222	.667	1.000
If $I_{ab} = 2 I_{be}$ (fairly stiff plate)	.167	.555	.833
If I_{ab} = infinity (absolutely stiff plate)	.111	.444	.667

The nail-plate junction

The nail-plate junction is an important region and often the site of failure. The junction should be designed so that there is no play between the parts. The stiffness of the junction should not be less than that of the strongest part of the appliance.

METHODS OF TESTING APPLIANCES

Earlier tests of appliances have been reported by Martz (1956), Foster (1959), and Frankel (1959). All of these tests were designed to show the amount of force, acting on the tip of the nail, that would produce gross plastic deformations in the appliance. Although reference was made to elastic deformations, no standard method of showing these changes was reported.

A standard test was designed, not only to aid in the study of the individual appliances under load conditions, but also to form some basis of comparison between various designs and materials. A knowledge of the behavior of the appliances under the test condition is necessary.

The deflection or downward bending that a given force acting on the tip of the appliance will produce can be measured. This deflection can be compared to the force acting on the tip of the appliance. To compare different appliances in this manner, their geometry must be similar.

The deflection can also be compared to the bending moment developed at a standard section in the appliances. If this is done, appliances of different design and geometry can be compared directly as to the deflection when a certain bending moment is developed. The measurements that must be made if the bending moment at the reference section (M_{ref}) is to be found are shown in figure 36. M_{ref} is expressed in Kpmm and is equal to l times R, the force acting on the tip of the appliance.

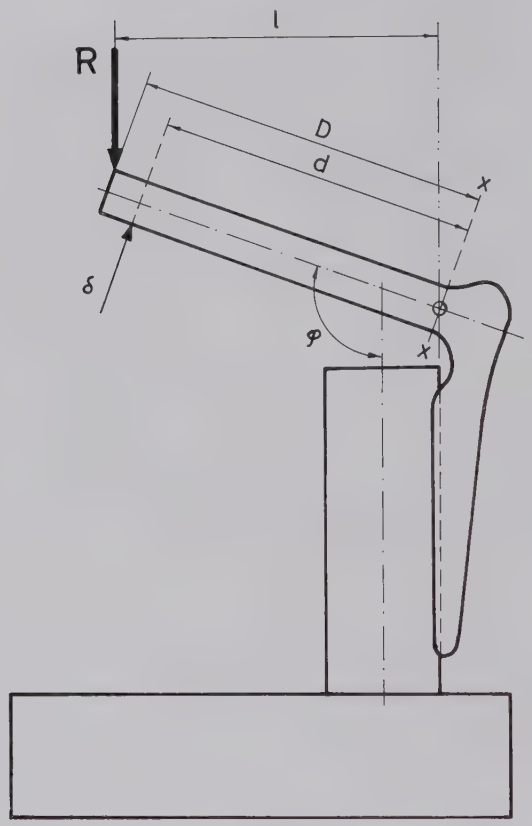


Fig. 36. Load deflection test geometry. x–x, reference section; R, force acting parallel to the plate on the tip of the nail; δ, deflection perpendicular to nail axis at a distance (d) from reference section; D, penetration depth of the nail; l, distance between center of gravity of reference section and action line of R; φ, nail-plate angle.

An apparatus for testing appliances was constructed of steel. It consisted of an upright post, to which the plate of the appliance could be bolted, welded to a base plate. An adjustable arm holding a Haldex extensometer was attached to the post. The extensometer could be read to 1/100 mm.

The appliance to be tested was firmly bolted to the post and measurements of the geometry made. The extensometer piston was arranged so that it was perpendicular to the long axis of the nail or screw (fig. 37). The test stand was then placed in the Amsler testing machine with the base plate resting on two steel rollers, which insured that the direction of the force would be constant.

Force was applied to the tip of the nail in gradually increasing increments, with

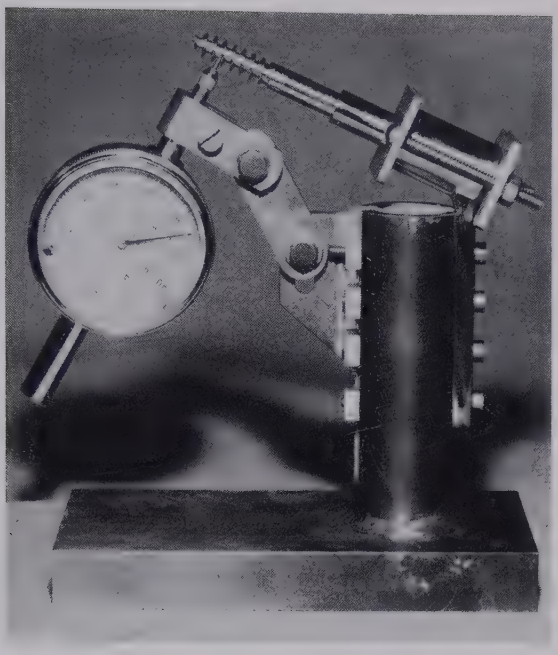


Fig. 37. The test stand. A Charnley compression screw-plate is mounted and ready for testing. The extensometer is adjusted so that the piston is at right angles to the long axis of the screw. It is necessary to place a clamp on the appliance to prevent the screw from backing out.

frequent decreases of force to the unloaded state. Measurements of the deflection were recorded. Graphs were constructed in which deflection was plotted against bending moment. From the graph shown in figure 38, it is possible to determine the elastic and plastic deformations found when five appliances were tested. The geometry of the appliances is found in Table 6.

TABLE 6. *Geometry of appliances.*

Appliance	Material	φ	l (mm)	D (mm)	d (mm)
Charnley	Steel	120°	76	90	87
McLaughlin	Vitallium	117°	74	87	86
Neufeld	Vitallium	135°	58	86	76
Jewett	Steel	135°	58	86	78
McKee	Steel	145°	40	69	50
Series 1–130	Steel	130°	61	84	78
Series 2 T	Titanium alloy	120°	78	93	85
Series 2 S	Steel	120°	78	93	85

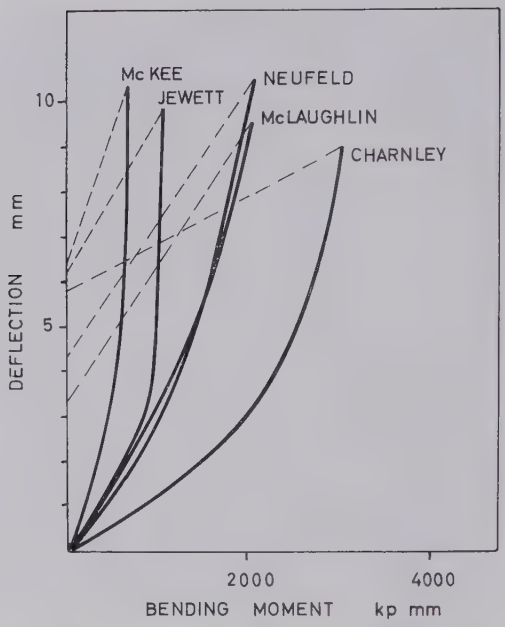


Fig. 38. Load-deflection diagrams. Solid lines, loading; interrupted lines, unloading. By drawing lines parallel to the unloading line which intersect the loading line, the amount of plastic deformity produced for any given bending moment can be found.

DESCRIPTION OF TEST NAIL-PLATES

From the results of the preliminary experiments reported in Chapter 5 and the data found in the test stand studies, it was concluded that currently used appliances bend under loading, a contributing factor in the disruption of experimental osteosyntheses and crushing of the interior of the femoral neck. The head of the femur and the femoral cortex below the greater trochanter were strong structures in the experiments. A hypothesis was drawn: the strongest possible osteosynthesis would connect the strong head to the strong lateral cortex, bypassing the neck, as much as possible, as a supporting structure for the appliance. To meet these conditions, a stiff and strong appliance was thought to be necessary.

With these requirements as a basis for design, a series of nail-plates were constructed. As noted by Haboush (1954) and Bechtol, Ferguson, and Laing (1959), the 'I' beam, a common engineering structure since the birth of modern engineering, was a good method of distributing material for strength and stiffness. The nails constructed for test purposes were made in a modified 'I' beam shape. The plates varied as to size and angle. Three series of appliances were made.

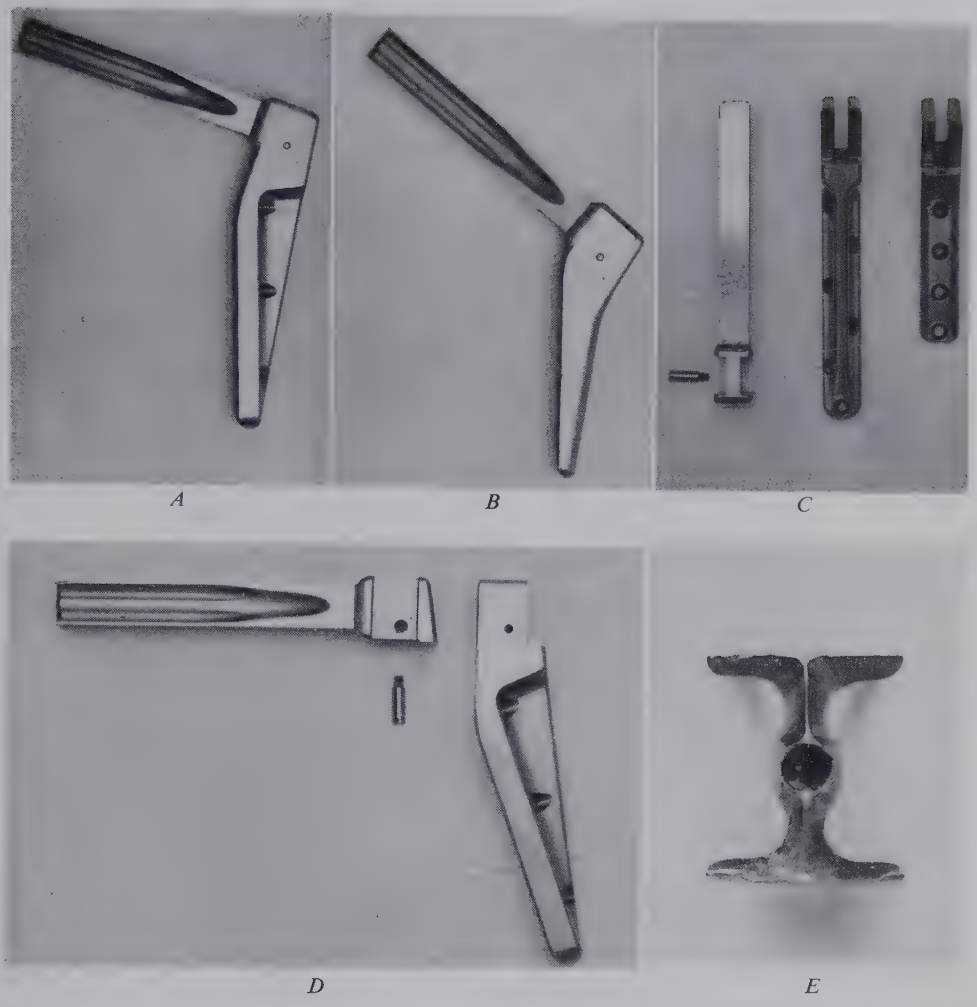


Fig. 39. Series 1 appliances. *A*, Nail-plate 1–110. *B*, Nail-plate 1–130. *C*, Series 1 components. *D*, Series 1 components. *E*, cross-section of Series 1 nail.

Series 1

These were made so that the load carrying ability of appliances of the largest possible size consistent with non-traumatic introduction into the femoral neck could be studied.

The appliance consisted of three parts: the nail, the plate, and a cross screw holding the two together. The angle of the nail-plate was not adjustable. Two plate

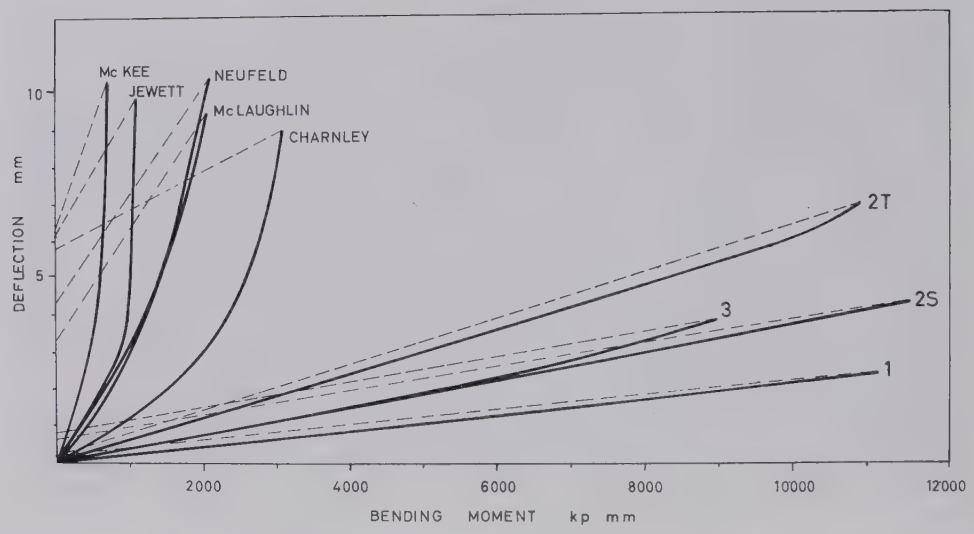


Fig. 40. Load deflection curves. 1, Appliance 1–130; 2 S, Series 2 (steel); 2 T, Series 2 (titanium alloy); 3, Series 3 (sliding). For explanation, see Fig. 38.

angles were made. Series 1–110 was made with a 110 degree nail-plate angle. Series 1–130 was made with a 130 degree angle (fig. 39).

The appliance was constructed of 17–14–4.5 strain hardened stainless steel. Geometrical data are found in Table 6.

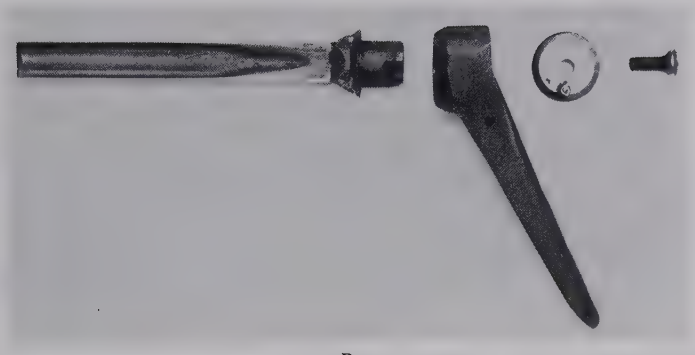
The bending moment at M_{ref} and the resultant elastic and plastic deformations from test stand studies are found in figure 40.

Series 2

This nail-plate had a nail with a similar modified 'I' beam shape, but the overall dimensions were smaller than those in Series 1, making the Series 2 nail-plate comparable to currently used appliances in size. It consisted of four parts: the nail, the plate, a locking pin, and a holding screw. The plate could be rotated on the end of the nail so that it could be adjusted to the femoral shaft after the nail was driven into the head (fig. 41). The nail-plate angle of 120 degrees was not adjustable.

In order to compare the characteristics of different metals with the same appliance design, nails were made both from 17–14–4.5 strain hardened stainless steel (Series 2 S) and titanium alloy (6 Al–4 V) (Series 2 T). Geometrical data are found in Table 6.

A



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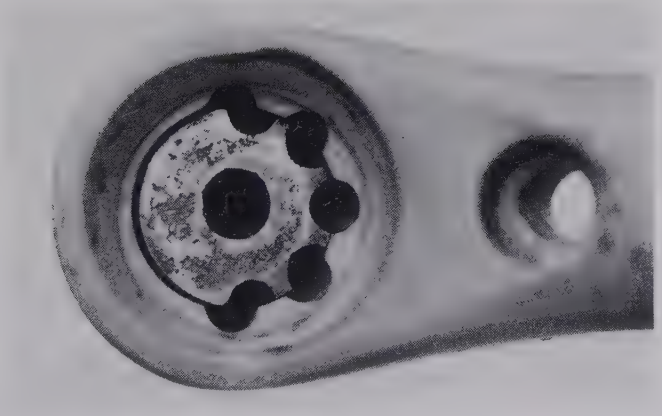


C

Fig. 41. Series 2 appliances. *A*, side view. *B*, components. *C*, nail-plate junction with mechanism to allow adjustment of plate to femoral shaft.

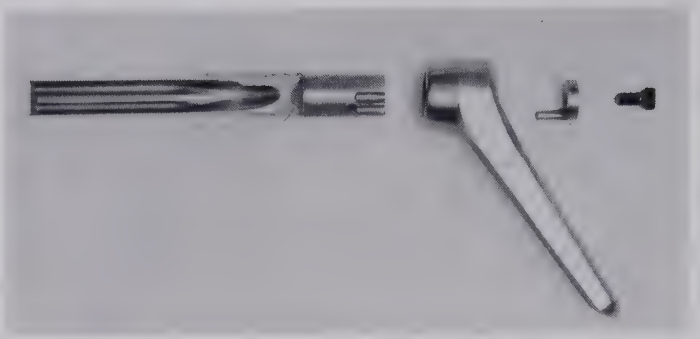


Fig. 42. Series 3 appliance.

Bending moment at M_{ref}, elastic and plastic deformation for both Series 2 S and 2 T are found in figure 40.

Series 3

Nail-plates with the same over-all dimensions as the Series 2 appliances were made, with the addition of a mechanism to allow extrusion of the nail through the plate as shortening occurred in the femoral head and neck. The appliance consisted of four parts similar to the Series 2 nail-plates (fig. 42). Construction was made of 17–14–4.5 strain hardened stainless steel. Geometrical data are found in Table 6.

As they had the same dimensions as the Series 2 nails except for the nail-plate angle of 130 degrees, bending moment, elastic and plastic deformations were found to be similar (fig. 40).

THE LOAD CARRYING CAPACITY OF APPLIANCES

With the data found in the load-deflection studies it was possible to compare different appliances as to load carrying ability, by selecting a standard amount of deflection and comparing the bending moments that produced the deflection. If appliance A and appliance B were compared, appliance B would have M_B / M_A times better load carrying capacity than appliance A.

The bending moments that were produced at M_{ref} when a deflection of 2 mm was observed is found in Table 7. By using Series 2 S as a standard, the relative load carrying capacities of the other appliances were determined.

Of the five appliances chosen to represent the types in common usage, the

TABLE 7. *The comparative load carrying capacity of appliances.*

(Deflection 2 mm).

Appliance	Actual Capacity (Kp/mm)	Relative Capacity (Series 2S = 1)
McKee	350	.06
Jewett	650	.12
McLaughlin	650	.12
Neufeld	750	.13
Charnley	1500	.27
Series 2T	3300	.59
Series 2S	5600	1.00
Series 3	5600	1.00
Series 1	9000	1.60

McKee and Jewett nail-plates had about the same load carrying capacity; this was also true of the McLaughlin and Neufeld appliances. The Charnley device had a greater load carrying capacity. The Series 1 appliances had a larger capacity compared to the other instruments. The Series 2 appliances had an intermediate position. The effect of varying the material with a similar design is seen in the difference in load carrying capacity between 2 S and 2 T, steel being much stiffer than titanium alloy.

CHAPTER 7

BEHAVIOR OF APPLIANCES IN OSTEOSYNTHESES

Total disruption has been used as the end point for the strength of an osteosynthesis in many of the studies reported previously. Investigators have studied the motion between the head and neck of the femur under load conditions. These methods have not given sufficient information of the behavior under less than destructive loads nor have they revealed the changes in the cancellous structures and the appliances.

The aim of the investigations to be described was to study osteosyntheses subjected to increasing loads. The following techniques were used to study the behavior of the appliances and the bony tissues:

The bending moment transported by the appliance to the shaft of the femur was measured with a strain gage technique.

The deflection of the appliance at a point near its tip was measured on serial roentgenograms.

The effect on the bone produced by loading the osteosynthesis was studied by sectioning representative specimens.

EXPERIMENTAL PROCEDURES

The specimens were, for the most part, those used in the determination of the ultimate strength of the femoral neck described in Chapter 4. Osteosynthesis was performed according to standard techniques. Specimens were mounted in the compression machine in the same fashion as during the fracture experiments. Vertical force was assured by the same arrangement of rollers and ball bearing shown in figure 11, page 38.

Strain Gage Studies

Strain gages manufactured by the Baldwin Locomotive Corp., Eddystone, Penna. were used. Their characteristics were: $k = 2.0 \pm 1.0\%$, $R = 120 \pm .3\ \Omega$. They were

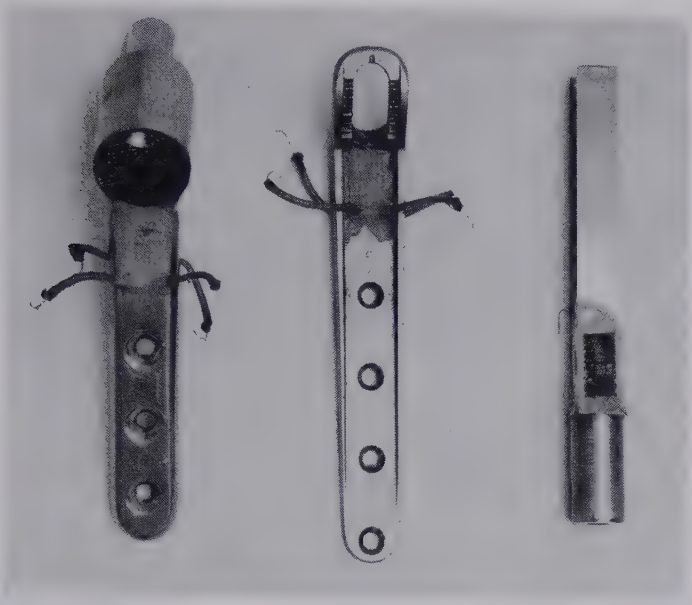


Fig. 43. Position of strain gages on appliances. (lt. to rt.) Charnley plate, McLaughlin plate, Series 3 nail (Similar to Series 2).

mounted on the appliances with glue supplied by the manufacturer (fig. 43). Waterproofing was performed by covering the lead wires with plastic tubing and additional glue. The strain gages were connected to the Philips bridge.

Calibration of the gages was performed by mounting the appliance on the test stand and loading and unloading the tip (fig. 44). Calibration curves were constructed for the various appliances, with the bridge reading plotted against bending moment at M_{ref}.

The information desired from the strain gage studies was the net moment transported by the nail to the plate and shaft of the femur through the nail-plate junction. Measurements of the moment transported could be made by mounting the gages at either the base of the nail or the top of the plate, as long as these areas were free from intimate contact with the bone. The strain gages were placed on the base of the nail in the Series 2 appliances; they were glued to the plates of the Charnley and McLaughlin instruments, the shape of the appliances making it impossible to place them elsewhere (fig. 43).

Static loads were placed on the specimens and bridge readings taken at 25 or 50 Kp intervals. The bending moment transported from the head to the shaft via the appliance was found with the aid of the calibration curves.

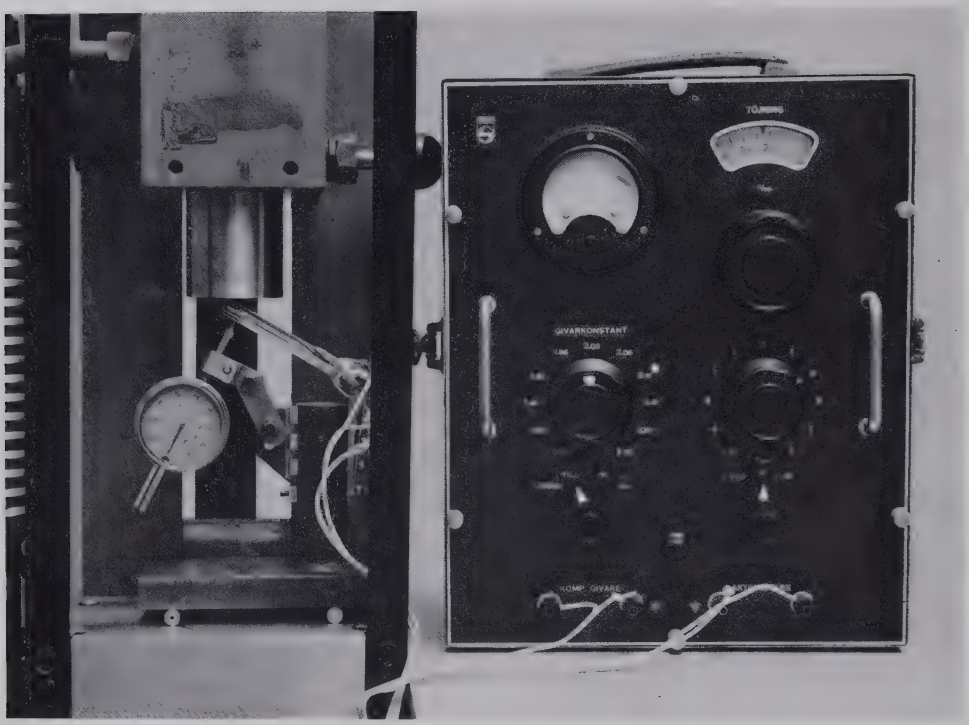


Fig. 44. Calibration of strain gages. The plate is bolted to the test stand, which is placed in the compression machine on rollers.

Serial Roentgenographic Studies

The roentgen apparatus and the film cassette were mounted in the same relationship to the bone described in Chapter 3. Exposures were made during the various stages of loading the specimen. Measurements of the deflection of the appliance were made by superimposing the films and noting the deflection of the nail or screw at the same distance from the tip used in the test stand studies. Measurements were made to the nearest 0.5 mm.

Errors in measurement were thought to be of several types. Enlargement of the image averaged 4 ½ per cent. This was measured by comparing the observed size of the appliance to the actual size. Another source of error was rotation of the appliance with respect to the cassette so that a true lateral profile was not seen. Rotation was noted in only a few cases and never amounted to more than 5 degrees. There may have been a few small errors in judging the correct number when rounding off to 0.5 mm was performed.

Sectioning of Specimens

Following the test of an osteosynthesis, the appliance was removed and the bone sectioned into slices approximately 5 mm thick. The plane of the sections was perpendicular to the long axis of the nail or screw. The entire head and neck were included in the sections.

STUDIES OF SERIES 1 OSTEOSYNTHESES

These experiments were performed to study the effect on the stability of an osteosynthesis of large and rigid appliances. The size of these appliances was near the maximum suitable for insertion into the femoral neck. Some indication of the limitations of internal fixation, as far as static strength was concerned, was desired

Table 8. *The effects of load on osteosyntheses made with series 1 nail-plates.*

Specimen	Age and Sex	Fracture data			Test data			Observations
		Type	Angle	Force (Kp)	Θ	ψ	Force (Kp)	
Appliance 1–110								
17 R	58 ♂	Subcapital	45°	Sawed	0	45°	580	Displacement, head fractured
19 R	75 ♂	Transcervical	80°	1010	0	52°	650	Displacement, head fractured
28 L	59 ♀	Subcapital	65°	730	0	53°	300	Inferior cortex of neck fractured, displacement
21 R	56 ♂	Transcervical	100°	700	23°	32°	570	Displacement, head fractured
23 R	77 ♀	Subcapital with spike	90°	450	23°	34°	300	Impaction and displacement
24 L	78 ♂	Subcapital with spike	75°	625	21°	30°	475	Displacement and impaction
28 L	59 ♀	Subcapital	65°	730	23°	30°	300	Impaction
29 L	83 ♂	Subcapital with spike	50°	800	22°	36°	700	Impaction, shaft broke below plate
Appliance 1–130								
19 L	75 ♂	Transcervical	25°	970	20°	37°	650	Impaction, head fractured
21 L	56 ♂	Transcervical	110°	850	23°	35°	290	Impaction, head fractured
23 L	77 ♀	Subcapital	45°	550	23°	21°	670	Impaction, slight displacement
24 R	78 ♂	Subcapital	55°	600	21°	28°	450	Impaction, inferior cortex fractured by impacting head
28 R	59 ♀	Subcapital	70°	795	23°	26°	500	Impaction, inferior cortex fractured
29 R	83 ♂	Transcervical	105°	680	22°	37°	520	Impaction

Fig. 45. Specimen tested with 1–110 nail-plate. Displacement of the head on the neck has taken place.

from the studies. The effect of steep and horizontal nails, in the absence of large deformations of the appliance, was studied.

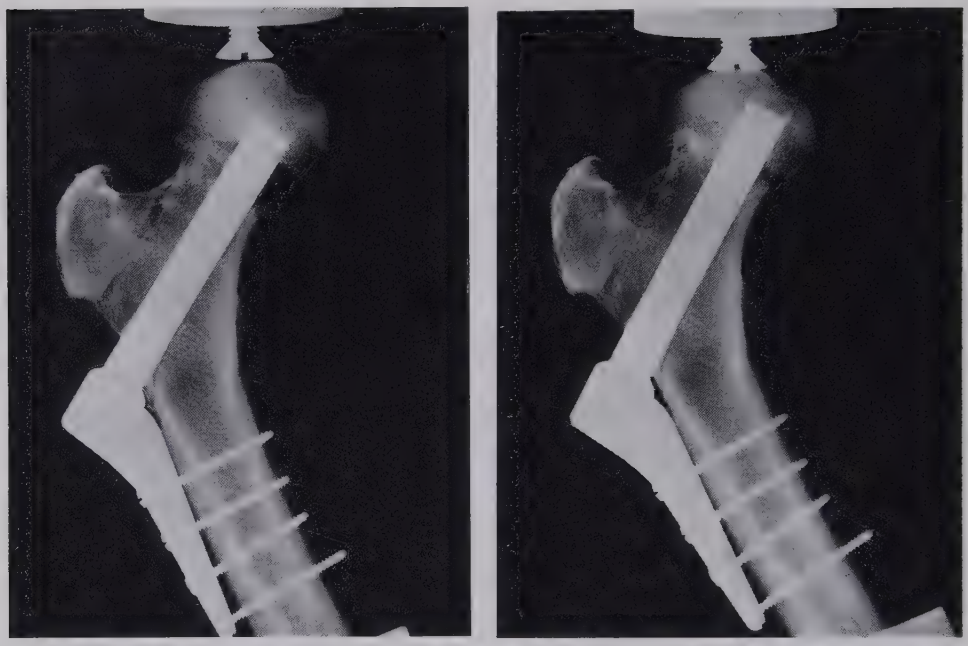
Data from the tests are found in Table 8. All specimens were inspected grossly. Tracings of the roentgenograms were used to study motion occurring between the head and the neck.

Osteosyntheses performed with nail-plate 1–110 were tested with the shaft of the femur held vertically (angle Ψ – angle u). Displacement of the head on the neck took place in all three tests (fig. 45). The head was fractured in two of these tests, and the inferior cortex of the neck was broken in the third. In the latter case, the nail was resting directly on the cortical shell. Although deflection of the nail was slight, as seen on serial roentgenograms and cross sections of the neck, it was sufficient to break the bone (fig. 46).

Tests were performed on nail-plate 1–110 osteosyntheses with the shaft of the femur tilted, in order to have the reacting force assume a direction consistent with that found during one leg stance. Combinations of impaction and displacement occurred, displacement usually predominant. The head was fractured in one test (fig. 47). The inferior cortex was not broken during any test.

Fig. 46. Roentgenogram of cross-sections of femoral head and neck tested with 1–110 nail-plate. Note slight crack in inferior cortex of neck. The outline of the appliance is sharply defined in the neck showing that negligible downward deflection has taken place; however, it was sufficient to fracture the bone. The head sections are intact.

Fig. 47. Roentgenogram of cross-sections of head and neck tested with 1–110 nail-plate. The head is crushed, the cancellous bone of the neck is intact. Reproduced through the courtesy of the Editor of Acta Chirurgica Scandinavica.



A *B*

Fig. 48. Specimen tested with 1–130 nail-plate. *A*, before test. *B*, during test. Impaction of the head on the neck has occurred.

Specimens with nail-plate 1–130 were tested in a similar fashion, with the femoral shaft tilted. The head was fractured in two specimens. Impacting took place in nearly all specimens (fig. 48). The impacting heads cracked the inferior cortex in two cases, the nails were not resting on the bone in either case.

Conclusions

Forces as high as 700 Kp were supported when the large nail-plates were used in an osteosynthesis. Disruption occurred in some cases when the head was crushed. In other cases, failure occurred when the inferior cortex of the neck was fractured either by the impacting head or the overlying nail.

Displacement and impaction occurred when an appliance with a 110 degree angle was used. With the 130 degree nail-plate, impaction usually occurred. It was not necessary to have an angle steeper than 130 degrees to produce impaction when a stiff nail-plate was used in an osteosynthesis tested with the force line acting in the same direction found during single leg stance.

STUDIES OF McLAUGHLIN, CHARNLEY, AND SERIES 2 OSTEOSYNTHESES

Table 9 contains data concerning the fracture types, appliances, and directions of force used in the tests.

Strain Gage Measurements

From the bridge readings and the calibration curves, the bending moment transported through the appliance to the shaft of the femur was found. These data are shown graphically in figures 49, 50, 51, and 52. Bending moment transported by the appliance is plotted against the load on the femoral head.

Serial Roentgenographic Studies

The downward deflections of the tip of the appliances for various loads on the femoral head are found in Table 9. In addition to the observed deflections, calculated deflections are presented. These were derived from the bending moment data found in the previous experiment and from the test stand studies (fig. 40).

It was noted that the observed deflections were usually larger than the expected deflections. There were several reasons for this discrepancy. The magnification factor of 4.5 per cent accounted for a small part of the difference.

There were differences in loading conditions in the test stand and in the osteosynthesis. In the test stand, force was applied to the tip of the appliance; no support was present under the nail or screw. In the osteosynthesis, the nail or screw was not loaded at the tip, but over an area extending from the tip to the fracture line. In addition, it received varying support from the bone. If support from the bone was taken into account, the observed deflection would be expected to be larger than the expected deflection. Figure 53 shows the bending moment along the course of a nail in the test stand and in a specimen.

Evidence for this theory was seen in the specimens with McLaughlin nail-plates, where the nails bent much more than would be expected from the bending moments transported. Roentgenograms of specimen 44 R are seen in figure 54. The observed deflection at 200 Kp was 2 mm; the expected deflection was only 0.5 mm. At this load, only 150 Kpmm were being transported to the femoral shaft by the appliance. A bend in the nail occurred at the site of support and in the area expected from figure 53.

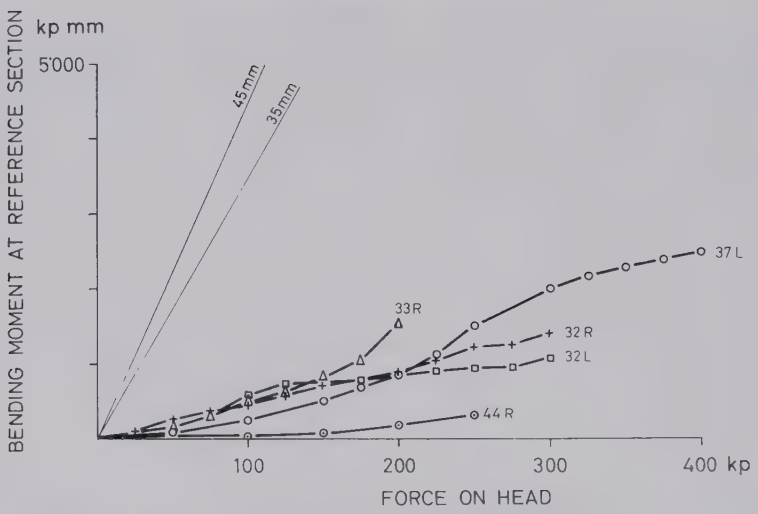


Fig. 49. Load-deflection curves for specimens tested with McLaughlin nail-plate. The interrupted lines show the total theoretical bending moments to be transported to the shaft at a reference point. The number in mm refers to the distance between the force action line and the reference point. By using the appropriate value for this distance (l_b) (found in Table 9), it is possible to find the amount of bending moment transported by the bone by subtracting the bending moment transported by the appliance from the total theoretical bending moment to be transported.

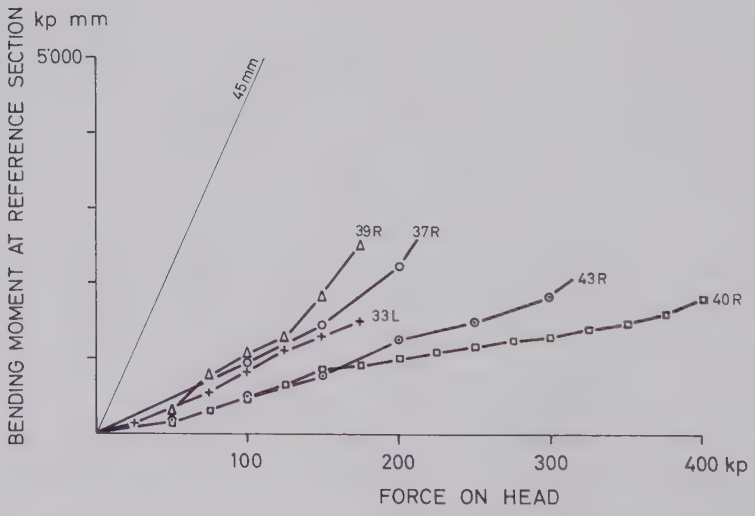


Fig. 50. Load-deflection curves for specimens tested with Charnley screw-plate. See Fig. 49 for description.

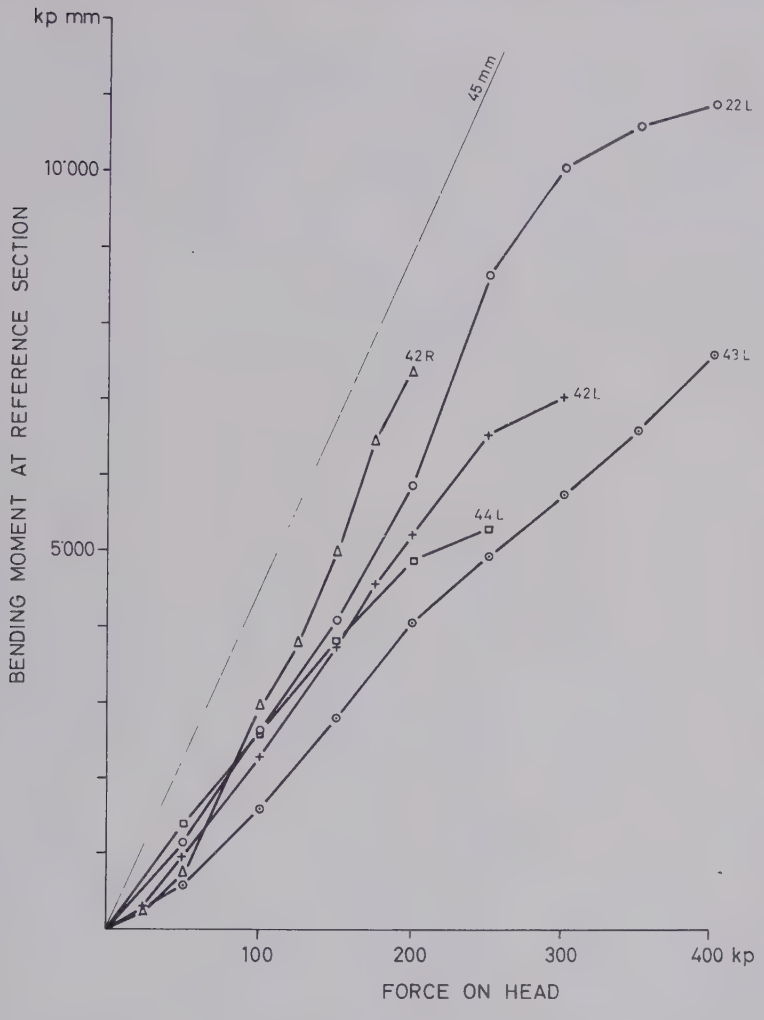


Fig. 51. Load-deflection curves for specimens tested with nail-plate 2 S. See Fig. 49 for description.

The Theoretical Total Bending Moment Transported to the Shaft

The theoretical total bending moment, M_b, is the bending moment at a point coincidental with the reference section of the appliance (corresponding to M_{ref}). Part of this total bending moment was transported by the appliance and has been demonstrated above. Part of the bending moment was transmitted by the bone. The total bending moment was found from the formula:

TABLE 9. *Behavior of McLaughlin, Charnley*

Specimen	Age and Sex	Fracture Data			Test Data					
		Type	Angle	Force	Appliance	Θ	ψ	L_B (mm)	Comments
32 L	81 ♂	Subcapital	50°	1000 Kp	McLaughlin	122°	23°	23°	42	Plastic deformation of nail
32 R	81 ♂	Subcapital	60°	1750	McLaughlin	135°	23°	27°	36	Plastic deformation of nail
33 R	50 ♀	Subcapital	50°	500	McLaughlin	132°	24°	26°	33	Plastic deformation of nail
37 L	77 ♀	Transcerv.	90°	750	McLaughlin	138°	23°	23°	28	Plastic deformation of nail plate junction
44 R	72 ♂	Subcapital	75°	1060	McLaughlin	140°	21°	29°	32	Plastic deformation of nail plate, metal creeping above 300 Kp
33 L	50 ♀	Subcapital	50°	490	Charnley	120°	24°	27°	37	Head tilted into varus abov 165 Kp, 4 mm extrusion
37 R	77 ♀	Subcap. + Spike	90°	1000	Charnley	120°	23°	19°	38	Creeping above 250 Kp, 3 mm extrusion
39 R	80 ♀	Transcerv.	90°	440	Charnley	120°	25°	28°	46	Gross slip of head on neck 165 Kp, 5 mm extrusion
40 R	70 ♂	Transcerv.	95°	975	Charnley	120°	23°	24°	54	Creeping above 400 Kp, no extrusion
43 R	82 ♀	Subcapital	50°	480	Charnley	120°	22°	27°	40	Creeping above 350 Kp, 5 mm extrusion
22 R	81 ♂	Transcerv.	90°		Series 2S	120°	21°	30°	50	Nail deformed above 19,000 Kpmm
42 R	75 ♀	Transcerv.	90°	470	Series 2S	120°	22°	23°	42	Lowest screw pulling ou 200 Kp
42 L	75 ♀	Subcapital	60°	620	Series 2S	120°	22°	28°	38	
43 L	82 ♀	Subcapital	45°	590	Series 2S	120°	22°	28°	31	
44 L	72 ♂	Subcapital	90°	1000	Series 2S	120°	21°	29°	49	
34 L	86 ♀	Transcerv.	85°	740	Series 2T	120°	20°	31°	48	Slight contact between hea and neck
35 L	42 ♀	Subcapital	40°	720	Series 2T	120°	17°	29°	42	Fracture reduced in sligh varus
35 R	42 ♀	Subcapital	45°	770	Series 2T	120°	17°	31°	39	Fracture reduced in sligh valgus
40 L	70 ♂	Transcerv.	65°	980	Series 2T	120°	22°	26°	42	

and Series 2 appliances in osteosyntheses

	Observed and calculated deflections for various loads on femoral head (in millimeters)											
	50 Kp		100 Kp		200 Kp		300 Kp		400 Kp		0 Kp (plastic deformation in bone)	
Specimen	Obs.	Calc.	Obs.	Calc.	Obs.	Calc.	Obs.	Calc.	Obs.	Calc.	Obs.	Calc.
32 L							7.0	3.5				
32 R							5.0	3.5				
33 R					6.5	6.2						
37 L			1.0	0.5	2.0	1.5	8	8.5	12	12	5	4.5
44 R	0	0	0.5	0.3	2.0	0.5	7 ~ 16				13	
33 L	0.5	0.4	1.5	1.1	4.0	> 2.5						
37 R			1.0	1.2	3	3.2	10	> 10			5.0	~ 5
39 R	0	0.2	1	0.6	1.5	1.3	2.5	1.7	4	> 2.5	3.5	> 3
40 R	1	0.4	2	1.5	5.5 (175 Kp)	4.5						
43 R	0.5	0.2	1	0.6	2	1.7	3	2.5	4	4.5	3	~ 2.5
22 R							7 (350 Kp)	4			6	> 5
42 R	0.5	0.3	1.2	1	2 (150 Kp)	1.8						
42 L	0.5	0.4	1	0.8	2	1.8						
43 L	0.5	0.2	1	0.6	1.5	1.4	2.5	1.8	4	2.8	~ 0	0.1
44 L	0.5	0.5	1	1	2	1.7	3	3	5.5	5	1	1
34 L			3	2.7								
35 L			1.5	1	2.5	2.3	4	3.3				
35 R			1	0.6	1.5	1.2	2.5	2.4				
40 L	1	0.8	2.0	1.6	4	3.4	6	5.2	8	7.5	0.5	0.4

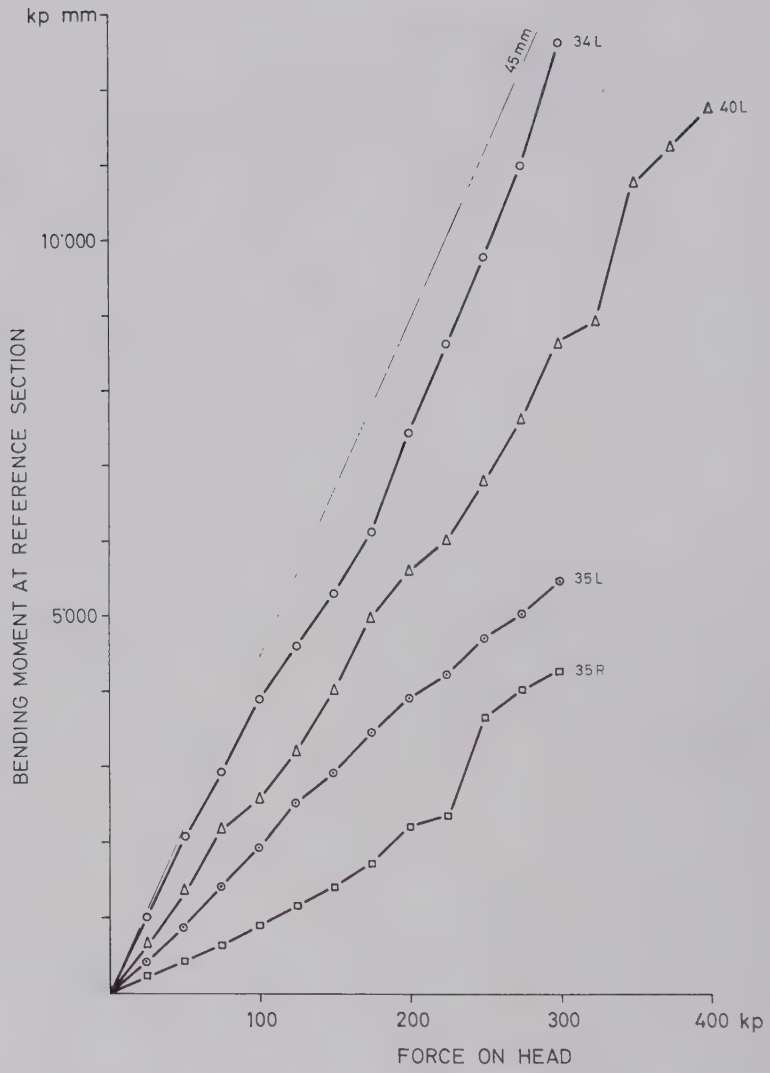


Fig. 52. Load-deflection curves for specimens tested with nail-plate 2 T. See Fig. 49 for description.

Equation 12 $$M_b = 1_b \cdot R$$

1_b is the distance between the force line and M_b
R is the force on the femoral head.

Figure 55 illustrates M_b and l_b. l_b was the subject of a magnification error averaging 4.5 per cent. Corrected values for l_b are found in Table 9.

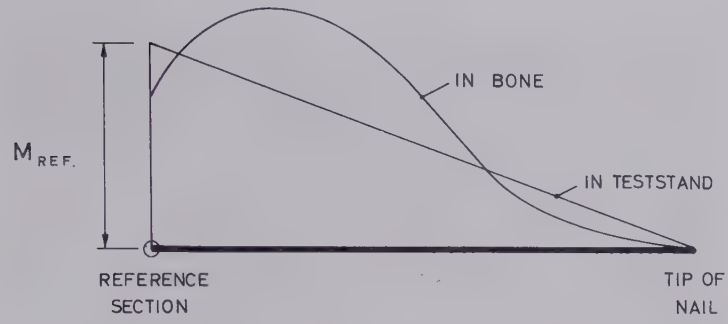
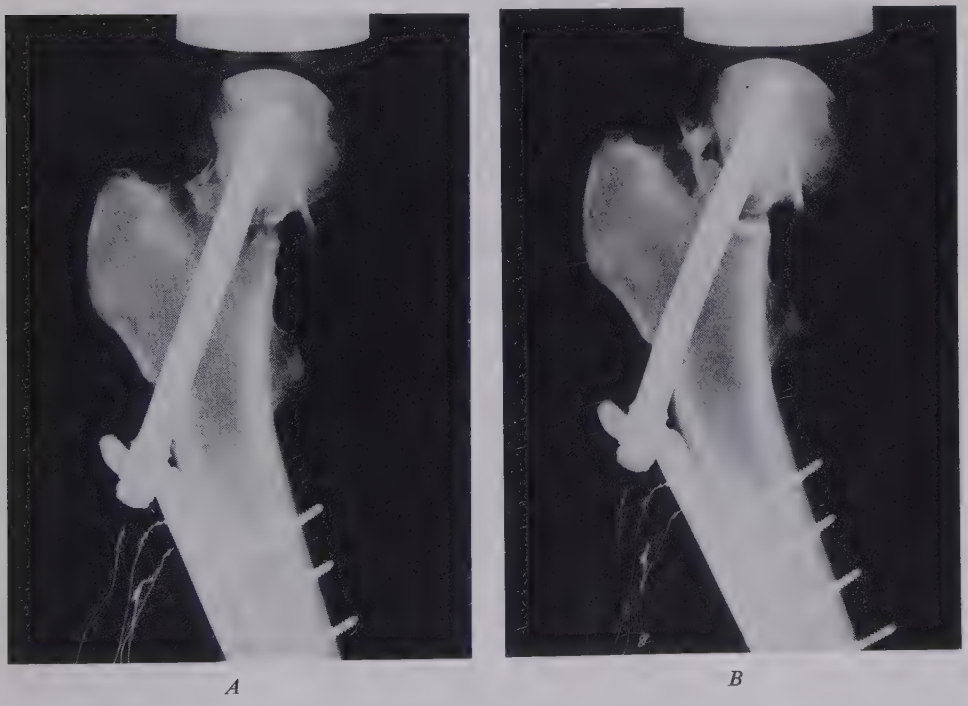


Fig. 53. Theoretical bending moments on a nail in the test stand and in an osteosynthesis (diagrammatic). When support from the bone is present, bending moment is higher along the nail, and may be lower at the reference section.



A *B*

Fig. 54. Roentgenograms of specimen tested with the McLaughlin nail-plate. *A*, before test. *B*, during test. Note bending of nail at point where it has received support from the bone. See Fig. 53.

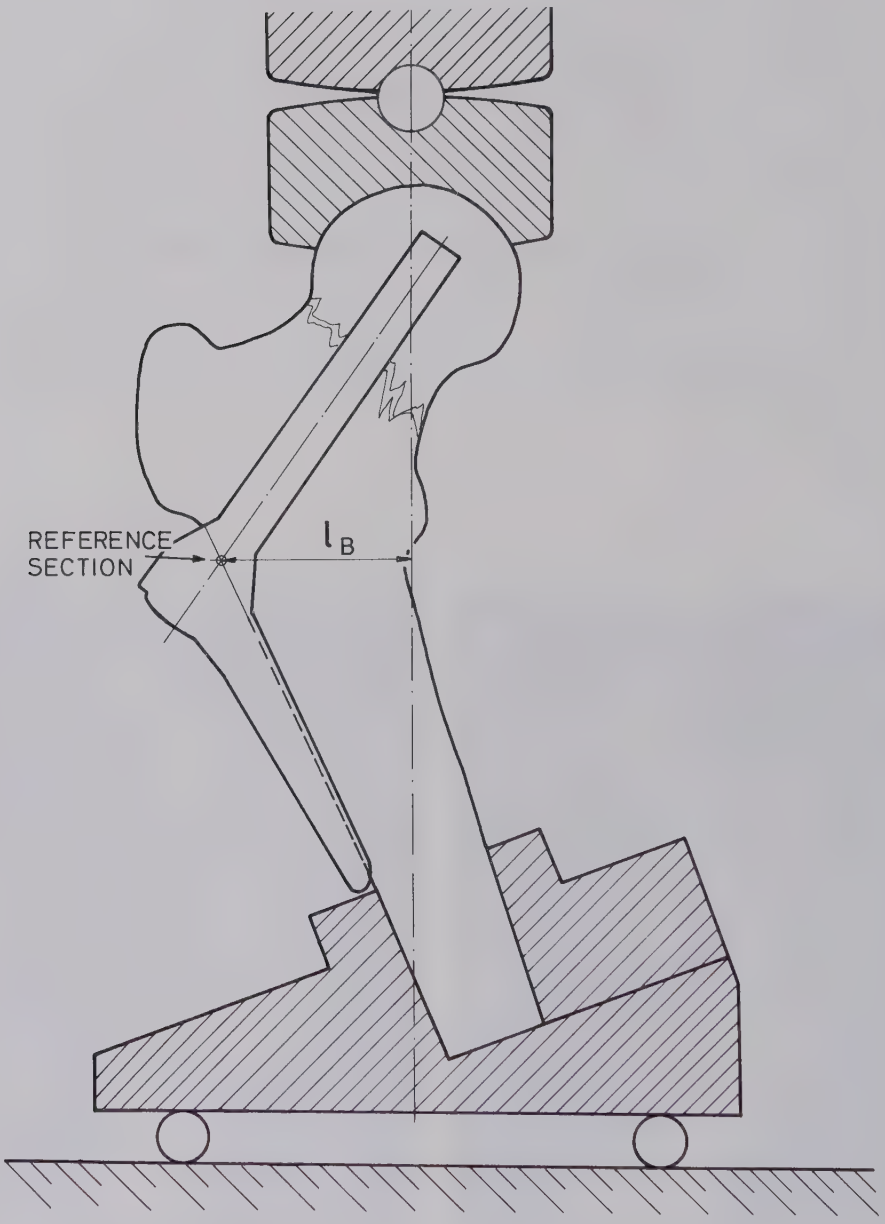


Fig. 55. The total theoretical bending moment to be transported from the femoral head to the shaft. l_b, the distance between the force line and the reference section of the bone. The point chosen as a reference point for the bone is coincidental with M_{ref} of the appliance.

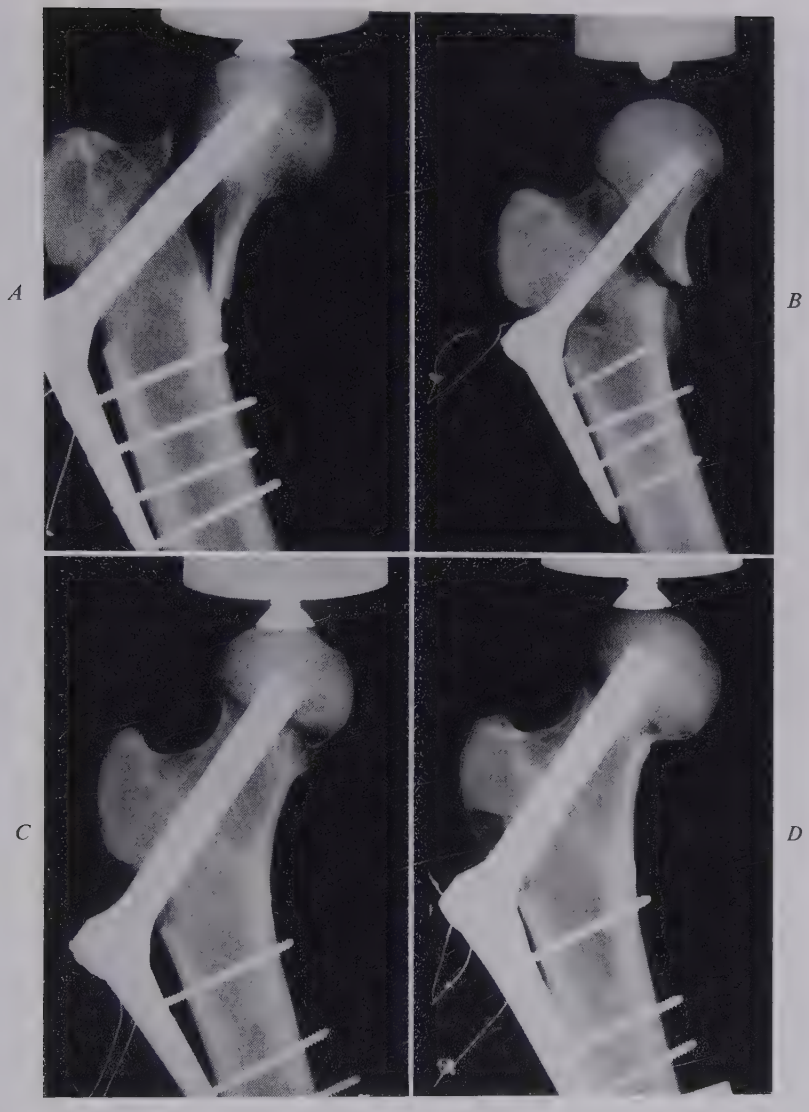
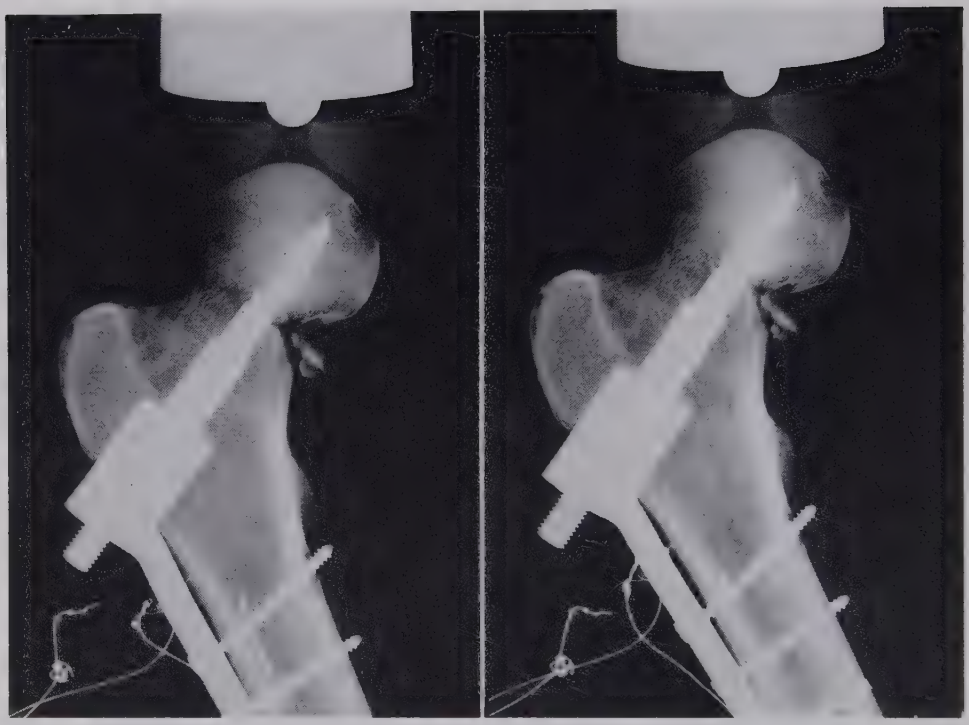


Fig. 56. Roentgenograms of specimens tested with appliance 2 T. Compare with Fig. 52. *A*, specimen 34 L. There is virtually no contact of the head with the neck. Only a small proportion of the total bending moment would be expected to be transported by the bone. M_{ref}, experimentally determined is very close to the line in Fig. 52 indicating the total theoretical bending moment to be transported. *B*, specimen 40 L. The fracture line is steep but there is more contact between the head and neck than in specimen 34 L. The bending moment transported by the bone is higher, as expected. *C*, specimen 35 R. *D*, specimen 35 L. The former specimen was nailed in slight valgus, and the latter specimen in slight varus. The fracture lines were otherwise similar. The specimen nailed in valgus had better contact between the head and neck. The bending moment transported by the bone is higher, as expected, in the valgus fracture as compared to the fracture nailed in slight varus.



A *B*

Fig. 57. Roentgenograms of specimen tested with Charnley appliance. *A*, without spring loading. *B*, with spring loading. Note impaction of head on neck in *B*.

M_b has been indicated in figures 50, 51, and 52 for l_b equal to 45 mm. In figure 49, two values for M_b are given: l_b equal to 35 and 45 mm.

For any given load on the femoral head, the part of the bending moment transported by the bone can be found by subtracting the bending moment transported by the appliance from a suitable value for the total theoretical bending moment.

Evidence in support of the above theory was found by studying the Series 2 T specimens. Specimen 34 L (fig. 56 *A*) had a vertical fracture line and was tested without impaction. There was virtually no contact between the head and the neck. The bending moment transported by the appliance is about the same as the total theoretical bending moment. Considering the lack of contact at the fracture site, it is not surprising that the bending moment transmitted by the bone is negligible.

The specimens shown in figure 56 *B*, *C*, and *D*, had fracture surfaces with greater contact. A greater part of the total bending moment was transported by the bone, a lesser part by the appliance. Specimens 35 R and 35 L were similar except that

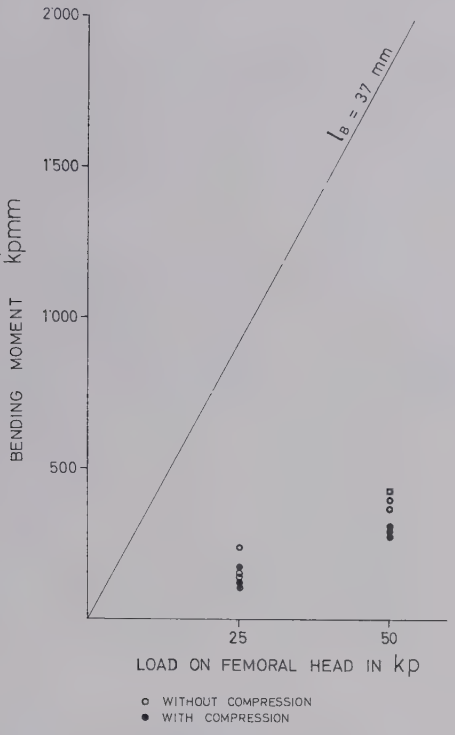


Fig. 58. Effect of compression on the bending moment transported by the bone and by the appliance. Solid dots, tests with compression. Open dots, tests without compression. The line marked l_b - 39 mm shows the total theoretical bending moment to be transported. See Fig. 55. In this experiment, compression lowers the bending moment transported by the nail about 25 per cent when a load of 50 Kp is acting on the head. Compression increases the part of the total bending moment transported by the bone by less than 6 per cent.

the former was nailed in valgus and the latter in varus. The specimen with the valgus reduction, with better contact between the head and neck, had more of the total bending moment transported through the bone. The specimen in slight varus had more of the total theoretical bending moment transported by the appliance.

l_b depends on the position of the line of force acting on the femoral head. If it is more vertical, l_b will be larger. During life, it depends upon the position of the femur and the resultant of the forces acting on the femoral head.

The Effect of Compression on Bending Moment Transportation

Compressive forces acting between the head and neck of the femur through the fracture line would be expected to alter the proportion of theoretical bending moment transported by the bone. The firmer the fracture surfaces were locked

together, the more they would act as a unit, with consequent lowering of the bending moment that would have to be transported by the appliance. Using a compression device, for instance of the Charnley type, loading experiments could be performed with and without compression.

Figure 57 shows the specimen used in an experiment in which loading was performed with and without spring compression. Figure 58 is a load-bending moment diagram of the experiment.

Compression lowered the bending moment transported by the appliance by about 25 per cent. If this change is compared to the total theoretical bending moment that must be transported, however, compression increased the part of the bending moment transported by the bone by less than 6 per cent.

The Comparative Load Carrying Capacity of Appliances in Bone

In Chapter 6, the comparative load carrying capacities of various appliances were determined in the test stand (Table 7). The load carrying capacity of appliances, expressed as bending moment transported (M_{ref}) was tabulated for an observed deflection of 2 mm. The relative load carrying capacities were compared to Series 2 S on the test stand and in the bone (Table 10).

Table 10. *The comparative load carrying capacity of appliances in osteosyntheses*

Appliance	M_{ref} (Kp/mm)	Relative load carrying capacity	
		test stand	osteosynthesis
Charnley	1.230 (1.100–1.450)	0.27	0.22
2 *T*	3.050 (2.600–3.650)	0.59	0.55
2 *S*	5.625 (4.900–7.400)	1.00	1.00

Specimen 34 L was omitted from the 2 T average.

Although there are many individual variations in the specimens and the direction of the fracture lines, there is good agreement between the test stand studies and the osteosynthesis tests. The supporting effect of the bone on the appliance accounts, in part, for the discrepancy between the test stand and osteosynthesis values. A small part of the difference in the figures for the Charnley screw-plate is due to the absence of the compressive factor in the test stand, and its presence in the osteosynthesis.

Conclusions

All appliances took up some load as soon as force was put on the femoral head.

When 50 Kp of load acted on the head, most specimens showed some bending of the appliance. The cancellous bone of the neck was compressed or crushed starting at this load.

The stiffer appliances transmitted more of the bending moment to the shaft of the femur.

The more elastic appliances bent more readily in the bone than did the stiffer appliances.

The stiffer the appliance, the less effect did bony support have on the deflection and on the bending moment transported by the appliance.

The relative load carrying capacity of appliances was of the same order of magnitude in the osteosyntheses as on the test stand.

CHAPTER 8

METALLURGICAL CONSIDERATIONS

In choosing the type of metal to be used in constructing an appliance, the stresses and strains it is subjected to are a major consideration. In internal fixation of the femoral neck, the appliances are subjected mainly to bending stresses. This bending stress acts on the nail, the nail-plate junction, and the plate.

It was shown that stiffer nail-plates transmit greater proportions of the total load on the head than more elastic appliances. With the distribution of bending stresses produced through support of the appliance by the bone, nails were plastically deformed. By choosing suitable materials and design, these elastic and plastic deformations might be avoided for loads presumed to be acting during single leg stance.

A large limitation on the metals available for use is that they must be non-corrosive in the body. They must have the ability to remain in the body for long periods of time and to be subjected to repeated stresses without undergoing electrolytic dissociation to the extent that local effects on the bone and soft tissues are produced, or that corrosion fatigue develops.

Aside from clinical tests, the method used today to judge the corrosion resistant ability of metals is the anodic back electromotive force test (Clarke and Hickman 1953). A potential drop is measured in an electrolytic cell containing equine serum. The test has been used to screen metals. Those materials with an anodic back EMF of over 400 mV. are considered suitable for testing in the body.

A measure of stiffness is the modulus of elasticity (E), measured in Kp/mm^2. The higher the modulus of elasticity, the stiffer the material.

The yield strength (σ_y), also measured in Kp/mm^2, is a measure of the ability of the metal to resist stress without significant plastic changes.

Table 11 gives data for the metals that are currently available. Under steel, both annealed and strain hardened varieties are described. The latter process is used to increase the yield strength of the material, but does not affect the modulus of elasticity.

Table 12 contains the author's ranking of the various materials, drawn up from the data in Table 11. This was done, not for the making of comparisons and decid-

TABLE 11. *Physical properties of metals used in internal fixation.*

Metal	Anodic back emf mV		Stiffness Kp/mm²		Yield strength Kp/mm²	
Titanium	+3500	(1)	11,000	(2)	24	(2)
Titanium alloy (6A1 + 4V)	~670	(4)	11,000	(2)	96	(2)
Steel						
18–8 annealed	460–490	(3,4)	20,000	(7)	22	(7)
18–8 strain hardened	460–490	(3,4)	20,000	(7)	~80	(7)
17–14–4.5 annealed (~A.I.S.I- 317)	~530	(4)	20,000	(2)	28	(2)
17–14–4.5 strain hardened (~A.I.S.I. 317)	~520–540	(4)	20,000	(5)	84	(5)
Vitallium	+650	(1)	25,300	(6)	~58	(6)

Key to numbers in parentheses

(1) Gillis (1958)
(2) Avesta Jernwerk (1958)
(3) Bechtol, Ferguson, and Laing (1959)
(4) Trägårdh (1959)
(5) Jarfall (1959)
(6) Hoyt (1952)
(7) Metals Handbook (1954)

ing which metal is best, but to point out that no perfect metal exists for the construction of appliances for internal fixation of the femoral neck.

All of the metals in Tables 11 and 12 are considered to have back EMF values above the level at which corrosion occurs grossly. Some of the metals, however, are more corrosion resistant than others.

TABLE 12. *Comparison of metals used for appliances.*

	Corrosion resistance	Stiffness	Yield strength
Titanium	1	3	4
Titanium alloy (6A1 + 4V)	2		1
Steel			
18–8 annealed	5	2	4
18–8 strain hardened 30%			2
17–14–4.5 annealed	4		3
17–14–4.5 strain hardened 30%			2
Vitallium	3	1	3

Fatigue strength is very important, but the present state of knowledge is not great enough to allow the fatigue resistance of test pieces to be compared to the fatigue resistance of the appliance when used in the body. The design is very important, as are the site and conditions under which the appliances are maintained in the bone.

The ideal material for appliances is corrosion resistant in the body, stiff, has a high yield strength, and is able to resist fatigue stressing.

CHAPTER 9

THE ANGLE OF INSERTION OF INTERNAL FIXATION APPLIANCES

When an appliance is used to stabilize a fracture of the proximal end of the femur, it will be subjected to various stresses by the load acting on the femoral head. The force acting on the head of the femur has two components acting on the appliance. One, the normal force, acts along the long axis of the nail. The other, the bending force, acts at right angles to the long axis of the nail, tending to bend the nail and to decrease the angle of the nail-plate junction.

The normal component of force will force the head onto the nail; the bending component tends to push the head into varus. The normal component may force the nail laterally out of the femoral cortex, unless it is held by a cross pin or a plate.

It was emphasized early in the modern period of internal fixation that steep nails were desirable both from an experimental and empirical standpoint. The steeper nails were thought to have a better grip on the lateral femoral cortex. Alignment of the nail in the principal load direction was believed to reduce bending stresses on the appliance.

One of the factors determining the amount of bending and normal force, is the direction of the reaction force on the femoral head. It is important to realize that this force direction may be variable. It is calculated as acting 10–15 degrees from the vertical, when standing on one leg. With a two leg stance it may be more vertical. At rest in bed, it may have other directions because of the action of different muscle groups.

It is conceivable, that under certain circumstances, the reacting force may be parallel, or nearly so, to the axis of the neck. The bending component on a steep nail would act, at least theoretically, in a direction tending to bend the nail and nail-plate junction upwards.

Since there is no single force reaction line, it is impossible to place an appliance so that it will always be in the direction of the reacting force.

Another determining factor for the bending and normal components of force on

the nail and the nail-plate junction, is the direction taken by the appliance in the femoral neck. There are two limitations to the range of values the appliance angle may have.

Experiments with the Series 1 appliances demonstrated that with stiff nail-plates, where the bending force on the nail produces negligible deflections, heads will be forced into varus if the nail-plate angle is 110 degrees. With appliance 1–130, impaction usually took place. When strong appliances were used, an angle of 120–130 degrees was usually sufficient to allow impaction, when the reacting force had a direction consistent with that produced during single leg stance.

The upper limit for the nail-plate angle is about 150–155 degrees, and depends on the fact that the steeper the nail, the higher in the head it must end. As the strongest part of the head is in the center, it would seem to be advantageous to place the tip of the appliance there.

The normal and bending components of the force acting on the appliance will in part be determined by the reacting force on the femoral head and in part by the steepness of the appliance.

Experimental data have been used to show that the lower limit for the nail-plate angle is about 120–130 degrees. The upper limitation is imposed by the anatomy of the proximal end of the femur and will be about 150–155 degrees.

Within these limitations, the effect of the bending force can best be overcome by the stiffness and strength of the appliance.

CHAPTER 10

SUMMARY

In Chapter 1, the history of internal fixation of the proximal end of the femur was reviewed with particular emphasis on the trend toward stronger fixation. Previous attempts to assess the efficiency of internal fixation were discussed. Pertinent studies on the structure and function of the upper femur were described.

In Chapter 2, a mechanical analysis of the forces acting on the head of the femur was presented. The importance of the ratio between the muscle and body weight lever arms to the magnitude of the force acting on the femoral head was demonstrated graphically. It was shown experimentally that forces acting between the greater trochanter and the lateral pelvic wall produce a rise in the force acting on the superior part of the femoral head, but not in the 'non-weight' bearing area. These studies demonstrated how the force on the femoral head depends upon the magnitude, direction, and location of the forces involved in pelvic equilibrium.

In Chapter 3, the manner of transmission of forces from the head of the femur to the trochanteric area through the cortical shell of the neck was demonstrated by means of the Stresscoat technique and strain gage measurements, which showed tensile and compressive strains produced on the surface of the bone. By proper positioning of the femoral shaft, it was possible to reproduce the force vector acting on the femoral head during standing on one leg. With force directions thought to be present during single leg stance, the strains on the femoral neck were more evenly distributed than when purely vertical force directions were used.

The strain pattern showed that the superior cortex of the neck was not the site of large tensile strains during single leg stance; in fact, it could be the site of compressive strains. Compressive strains were found to be distributed in the inferior cortex, diminishing gradually in the anterior and posterior cortices. When the femur was positioned so that the force vector acting on the head was more vertical, the strain pattern was different. Tensile strains were present in the superior cortex. Compressive strains were present in the inferior cortex. There was a steeper gradient of strain distribution as the anterior and posterior cortices were approached, and these structures were found to be under very little strain.

In Chapter 4, determination of the ultimate strength of the femoral neck was

discussed. Using a force direction coincident with that found during single leg stance, femurs were loaded until fractured. The male specimens had a higher average breaking strength than did the female specimens.

The fractures were classified into three types according to the appearance and localization of the fracture line. The types were: subcapital, transcervical, and an intermediary type consisting of a subcapital fracture with a spike of neck attached.

There was no correlation between fracture type and ultimate strength.

A vectorial analysis of the force acting on the head and neck was made, and the normal and bending forces on the neck calculated. A correlation was found between the type of fracture and the direction of force. As the direction of force determined the proportion of normal to bending force, a correlation was also found between this ratio and the fracture type.

Subcapital fractures were produced with higher normal to bending force ratios, when the line of force approached the longitudinal cervical axis. The transcervical fractures were found with lower ratios, when the force direction was more perpendicular to the cervical axis. The intermediate types were found in the middle range of normal to bending force ratios.

Subcapital fractures occurred by compressive failure of the neck. Transcervical fractures occurred by tensile failure of the superior cortex of the neck.

With static loading, the type of fracture was only influenced by the direction of the force.

Preliminary studies of the mechanism of failure of osteosyntheses were reported in Chapter 5. Observations were made of the breakdown force, the plastic deformation of the appliance, and the resistance of the bone. Failure was due to bending of the appliance so that it rested on the inferior cortex of the neck and fractured that area.

The femoral heads were not crushed during the tests. Studies of isolated femoral heads mounted on four flanged nails showed that they resisted crushing by the nail better when the nail was more horizontally oriented. The femoral cortex below the trochanters had good screw holding ability.

A hypothesis was drawn that there were relatively weak and strong areas in an osteosynthesis. To have the strongest possible osteosynthesis, the force acting on the head of the femur should be transmitted mainly via the appliance to the femoral cortex.

In Chapter 6, the static strength of appliances was described from the theoretical standpoint. An idealized appliance was discussed and the importance of the stiffness of the nail and plate to the deflection produced by loading the tip was stressed.

A method of testing appliances was described. Deflection at the tip of the nail was correlated with bending moment at a standard reference section. The elastic

and plastic deformations produced by loading the tip of the appliances were illustrated.

Descriptions were given of test nail-plates, designed to study the effects of connecting the head to the shaft with a strong appliance. Several types were made. Series 1 appliances were of a large size and were stiff. They were used to demonstrate the effect of having an appliance which was as large as could be placed atraumatically in the neck.

Series 2 appliances were designed to study the load carrying ability of a nail-plate whose size was near to that of currently used appliances. By using two different metals, it was possible to show how the stiffness of the appliance was influenced by the modulus of elasticity of the material. An appliance made of steel was stiffer than one made of titanium alloy.

Series 3 appliances were designed to show that a sliding mechanism could be incorporated in a design without great change in the load carrying ability.

Graphs were presented which showed the elastic and plastic deformation, and the bending moment produced by loading the tip of the appliance.

By selecting a standard amount of deflection, 2 mm, the bending moments that could be transported by the various appliances were compared. Using appliance 2 S as a standard, the relative load carrying capacities of the various appliances was given.

The behavior of appliances in bone was discussed in Chapter 7. Studies with Series 1 nail-plates showed that with the shaft held vertically during loading, failure occurred owing to crushing of the head of the femur. Negligible deformation of the appliance took place. In one case, the inferior cortex was cracked by a nail which was in intimate contact with the bone.

When the force on the head of the femur had the same direction as was found during one leg stance, failure of the osteosynthesis occurred because of fracture of the head or cracking of the inferior cortex of the neck by the impacting head. Displacement of the head on the neck took place with nail-plate angles of 110 degrees. With the 130 degree appliance, impaction of the head on the neck predominated.

Studies on the strength of osteosyntheses made with McLaughlin nail-plates, Charnley compression screw-plates, and Series 2 nail-plates, were performed. Several techniques were used to study the course of events as the osteosynthesis was loaded. Force directions were consistent with single leg stance force vectors. The bending moment transported by the appliance to the shaft of the femur was measured by strain gage techniques. This bending moment was compared to a total theoretical bending moment developed by the external force. The latter bending moment was calculated from the known geometry and the external load.

The deflections of the appliances under loading were measured using a serial roentgenographic technique. Comparisons were made of the observed deflections and the deflections calculated from the bending moments. There was close agreement between the calculated and observed deflections if the roentgenographic magnification errors and the supporting effects of the bony tissue on the bending moment were taken into account.

The effect of compression between the head and neck was studied, and it was found to reduce the bending moment transported by the appliance by about 25 per cent. It increased the part of the total bending moment transported by the bone by less than 6 per cent.

The relative load carrying capacities of various appliances were compared on the test stand and in the bone. The appliances had relative load carrying capacities of the same order of magnitude, both in the bone and on the test stand.

It was concluded that the appliances transmitted some load as soon as force was put on the femoral head. The stiffer appliances transmitted more bending moment to the femoral shaft. The more elastic appliances bent more readily in the bone than did the stiff appliances. The stiffer the appliance, the less effect did bony support have on the bending moment transported and on the deflection of the appliance.

In Chapter 8, a consideration of the various metals that could be used in construction of appliances was made. Using standard engineering data, a table of comparison was drawn up of the various metals. It was concluded that from the standpoint of corrosiveness, yield strength, and stiffness, there was no perfect metal.

In Chapter 9, the angle of insertion of nails was discussed. The limitations on the range of angles for nail-plates were described. About 120–130 degrees was thought to be the lower limit for stiff appliances. The upper limit was considered to be 150–155 degrees, as higher angles would place the tip of the nail in the superior distal aspect of the head, where the cancellous bone is not as dense as in the center. Furthermore, it was concluded that it was impossible to place the nail in line with the reacting force on the femoral head, as this direction was not constant but depended on the type of stance and muscular action.

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